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Muscle mechanics in diagonal stride cross-country skiing and the effect of speed and incline

Abstract

Background: In gaits, such as running and walking, decoupling of the muscle fascicles length changes from the length changes of whole muscle-tendon unit (MTU), allows muscle fascicles to work almost isometrically and close to its optimal length, and therefore the fascicles can generate high forces economically. This is possible because the compliant tendons are stretch in the first part of the stance phase, facilitating storage of elastic strain energy within the elastic elements (EE), which, then is returned in the last part of stance. The storage and release of energy by the EE contributes substantially to the work done by the MTU during locomotion, thus decreasing the demand for muscle work. It has been speculated that a similar muscle-tendon dynamic to running takes place during a diagonal stride (DIA) in cross-country (XC) skiing, and that skiers store elastic energy in the preload phase prior to the kick. However, in vivo measurements of the fascicle length and contractions velocity, while the skiers perform diagonal stride, has not been conducted.

Purpose: The purpose of the study was to explore gastrocnemius medialis (GM) muscle-tendon mechanics during DIA on roller skies in vivo, and to examine the effects of different speeds and inclines on muscle-tendon behavior. Firstly, we hypothesize that propulsion in the kick phase is generated by concentric muscle contraction in the GM muscle while the MTU undergoes a stretch-shortening cycle. Furthermore, we believe that muscle contraction is initiated during the late part of the glide phase. Secondly, we hypothesize that the muscle-tendon behavior during the kick phase is affected by increase in speed and incline.

Methods: 13 elite senior XC skiers (age 26.5 ± 4.8 years) roller skied on a treadmill using diagonal stride (DIA) under three different conditions: (speed: 2.5 m s^{-1}, incline: $5°$) named DIA_{ref}, (speed: 3.5 m s^{-1}, incline: $5°$) named DIA_{fast}, and (speed: 2.5 m s^{-1}, incline: $10°$) named DIA_{steep}. Ultrasound measurements of the GM muscle fascicle length was synchronized with kinematic, kinetic, and muscle activity (EMG) measurements of the right leg. A single-sided One-Sample T-tests were used to test muscle-tendon unit (MTU) behavior in DIA_{ref}. One way repeated-measures ANOVAs were used to test how the increase in speed and incline affected the MTU behavior.

Results: There is a stretch shortening cycle of the MTU during the kick phase, with a significant stretch in the first part 9.91 ± 0.29 mm (P<.0001), followed by a significant shortening 24.29 ± 7.79 mm (P<.0001). The GM muscle fascicles shortened throughout the entire kick phase 9.1 ± 2.7 mm (P<.0001). There was a significant stretch of the MTU in the late glide phase, whilst there was no change in fascicle length. The GM muscle is significantly active during the kick phase, but not significantly active in the late glide phase.

Both increase in speed and incline increased MTU stretch (P<.0001). There was no significant difference in fascicle shortening velocity between DIA_{ref} and DIA_{fast}, whereas the fascicle shortening velocity decreased with increased incline (P<.0007). Integrated EMG (iEMG) was significantly higher in DIA_{steep} compared to DIA_{ref}, whereas DIA_{ref} and DIA_{fast} did not differ.

Conclusion: This study showed that during the kick phase of DIA XC-skiing, the GM muscle fascicle shortens continuously, whereas the MTU lengthens in the first part of the kick phase, allowing for storage of elastic energy, which is released later in the kick phase when the MTU shortens rapidly. We found no conclusive evidence for storage of elastic energy being initiated in the glide phase. Fascicle shortening velocity did not increase with higher speeds, however at increased incline the fascicle shortening velocity decreased. This is due to how the muscle-tendon interact under different time constraint and requirement for force and work. The increased demand at higher speeds is met by higher relative contribution of the tendons to the force generated (i.e., increased stretch and recoil of energy). The increased demand at increased incline is met by a longer kick phase, allowing fascicles to operate at lower shortening velocities and at a more favorable part of the force-velocity curve to produce higher forces.

Keywords: Muscle-tendon unit, ultrasound, biomechanics, cross-country skiing, diagonal stride

Preface

To write this master thesis has been a great and challenging experience, and as expected, it had its fair share of ups and downs. The completion of this thesis would not be possible without the help and guidance of my supervisors, friends, and family. I am forever grateful and would therefore like to share my gratitude. First and foremost, I would like to thank my main supervisor Amelie Werkhausen and my co-supervisor Øyvind Nøstdahl Gløersen for all their help and support in writing this master thesis. They have really gone above and beyond any measure, both as supervisors and as fellow humans.

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Anders Lundervold

Asker, June 2021

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Introduction

Cross country (XC) skiing is an Olympic winter sport, where athletes propel themselves through a groomed snow track using their skies and poles. XC-ski competitions are done in either the classical or the skate (free) technique. Both techniques consist of several sub-techniques, which act as a gearing system (Nilsson et al., 2004). Diagonal stride (DIA) is a sub-technique in classical skiing DIA, typically used in slower conditions such as uphill sections (Smith, 2003). When using this technique, the skier performs a kicking action on one leg followed by a weight shift to the contralateral leg where the skier glides prior to initiating a new kick. Initially DIA appears similar to running, but because of the glide phase in DIA, it may have different muscle-tendon mechanics (Kehler et al., 2014).

During walking and running it was found that, when the muscle is active and the elastic elements (EE) in the tendon and aponeurosis is lengthened, a considerable amount of elastic energy can be stored in the muscle-tendon unit (MTU), which can be utilized later in the kick phase (Roberts & Azizi, 2011). This could potentially increase energy conservation and power amplifications in the MTU, improving the performance in skiing competitions. The role of EE for energy conservation in running has been studied extensively (Bohm et al., 2019; Farris & Sawicki, 2012b; Lichtwark et al., 2007; Monte et al., 2020). They have found that utilization of elastic energy from the EE allows the muscle fascicles to work at lower shortening velocities, and at a more optimal level to produce high forces with less work required and thus higher efficiency.

From angular velocity curves and electromyography (EMG) during DIA XC-skiing (Komi & Norman, 1987) found that the muscle undergoes a stretch-shortening cycle while being under constant electrical stimulation. They theorized that because of the stored elastic energy, a given level of force could be produced with a lower motor unit activity. Thus, implying that a lower metabolic energy expenditure is required by the muscle. However, due to technological limitations at that time, they were limited to the use of electromyography (EMG) and two-dimensional (2D) angular kinematics, and they were only able make indirect inferences about the muscle fascicle and tendon behavior.

Kehler et al. (2014) found that during DIA, the center of mass (COM) is briefly lowered (i.e., the gravitational potential energy (GPE) decreases) prior to the initiation of the kick. Then

during the kick, the GPE and the kinetic energy (KE) increase concomitantly. Thus, (Kehler et al., 2014) speculated that some of the elastic energy in the EE might be stored during the pre-stretch in the preload phase during DIA. This may contribute to more favorable muscle-tendon dynamics.

To be able to say whether elastic energy is stored and then later utilized during DIA we need to study the fascicle shortening velocity and length changes in vivo. Prior research on the muscle-tendon mechanics during DIA XC-skiing has been limited to indirect estimates of the fascicle behavior only. Thus, the aim of this master thesis is to explore gastrocnemius muscle-tendon mechanics during diagonal stride on roller skies in vivo using ultrasound measurements, and to examine the effects different speeds and inclines have on muscle-tendon behavior.

The main hypothesis is that propulsion during the kick phase is generated by a concentric muscle contraction in the GM muscle while the MTU undergoes a stretch-shortening cycle. Furthermore, we hypothesize that muscle contraction is initiated during the glide phase, facilitating elastic energy storage in the elastic elements (EE) prior to the kick. This should be seen as a simultaneous EE lengthening and EMG activation during the late part of the glide phase. The second hypothesis is that muscle-tendon behavior during the kick phase changes with increased speed or incline. Firstly, we believe that with increased speed, the GM will show an increase in EMG activation, higher fascicle shortening velocities and an increased stretch of the EE, due to a shorter kick phase. Secondly, at an increased incline we expect to see higher EMG activation, a similar fascicle shortening velocity and an increased stretch of the EE during the kick phase, due to the increased force demand.

Background

This section aims to present the basic biomechanics of the musculoskeletal system. Firstly, tackling the morphological and material properties of the skeletal muscle and tendons separately. Followed by linking their function together and presenting some of the current knowledge about the muscle-tendon behavior during locomotion and its importance. Lastly, to look at the current knowledge of muscle-tendon behavior during DIA XC-skiing. This section focuses primary on the on the musculoskeletal biomechanics of striated pennate muscles, such as the gastrocnemius muscle investigated in this study.

Human skeletal muscles

Skeletal muscles are one of the most dynamic and plastic tissues in the human body. From a mechanical point of view the function of skeletal muscles is to convert chemical energy into mechanical energy and through this, generating force and power to produce movements or maintain posture. (Frontera & Ochala, 2015).

Muscle morphology

A single muscle cell is called a muscle fiber. The sarcomere is the smallest functional unit within the muscle fiber and responsible for its contractile properties. Within the sarcomere there are repeating actin and myosin protein filaments partly overlapping and connected by cross-bridges shaping the myofilaments, that in combination with other proteins the myofilaments form a myofibril. Muscle fibers are in turn built up by several hundred myofibrils in parallel and sarcomeres in series. Each muscle fiber is covered by a cell membrane called sarcolemma. External to sarcolemma is the endomysium, a connective tissue sheet that encases the muscle fiber and anchors it to other muscle fibers and connective tissue, and eventually to the tendon. Hundred or more muscle fibers are bundled together to form a muscle fascicle. Each fascicle is encased by the connective tissue sheet, the perimysium. Lastly the fascicles are bundled together by the epimysium to form the whole muscle (Enoka, 2015; McGinnis, 2013). The hierarchal structure of the muscle can be seen in Figure 1.
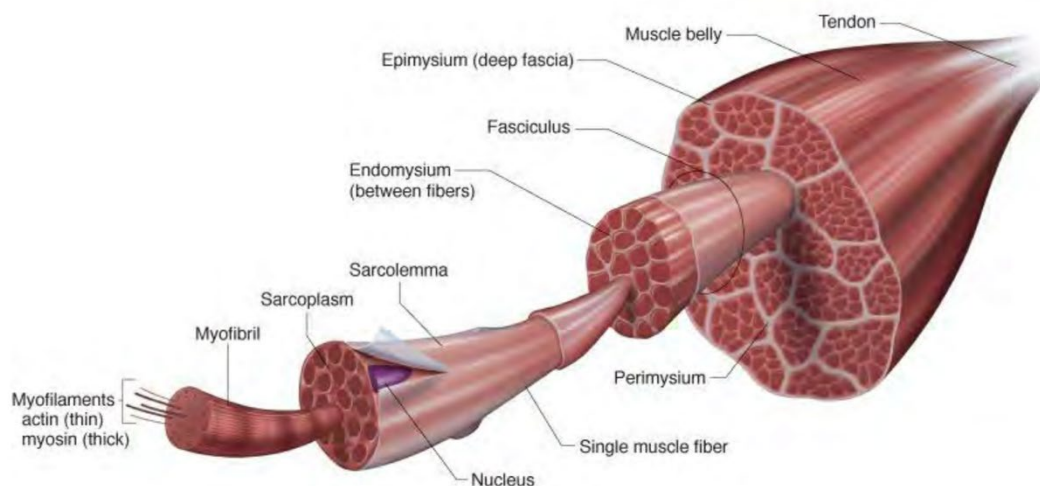
Figure 1 Skeletal muscle macrostructure from (McGinnis, 2013)

Mechanical properties

Muscle activation results in muscle actions that can be classified into three basic types: isometric, concentric or eccentric contraction. Concentric muscle action results in shortening of the muscle fascicles, and subsequently shortening of the whole muscle. Eccentric muscle action is when the muscle fascicles lengthen under load. Lastly, isometric contraction is when muscle fascicles contract generating tension and force, but the fascicles length and velocity remains relative unchanged and close to its original length.

Force generated by the muscle depends on many factors including the degree of activation by the nervous system (i.e., recruitment of motor unit), muscle architecture (including the pennation angle and cross-sectional area (CSA) described below). and the number of actin-myosin cross-bridges formed (Frontera & Ochala, 2015).

Due to variations in overlap of the actin and myosin filaments during contractions, muscle force productions are length dependent, which has been described as the force-length (F-L) relationship of muscles (Gordon et al., 1966) (Figure 2). Force-length of the muscles is influenced by the joint angles of the muscles, therefore there is an optimal length or joint angle where muscular force is maximized. GM muscle has been shown to operate at the ascending limb of the force length relationship in running (Arnold et al., 2013).
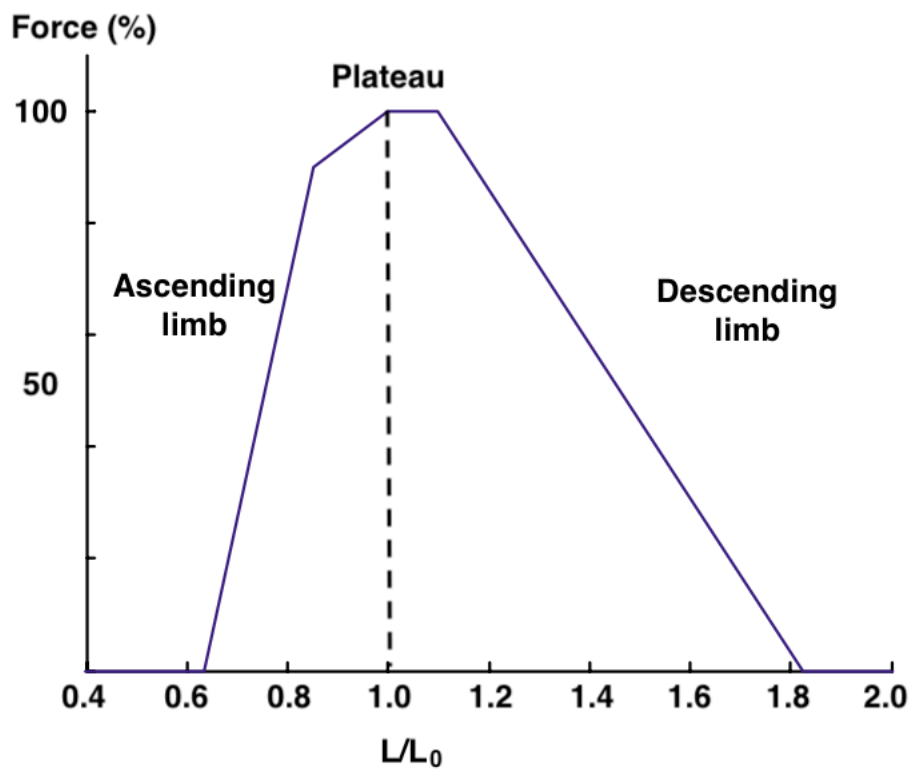
Figure 2 Sarcomere force -length relationship, first described by (Gordon et al., 1966), by scaling the number of cross-bridges possible

The force generation capability of muscles is also dependent on contraction velocity. The maximum force a muscle can generate, decreased with increased shortening velocity, and the maximum contraction velocity decreases with increasing force (Hill, 1938). Further, Hill later found that the power output of the muscle is maximized at approximately 30% of maximum shortening velocity (Hill, 1964) (Figure 3).
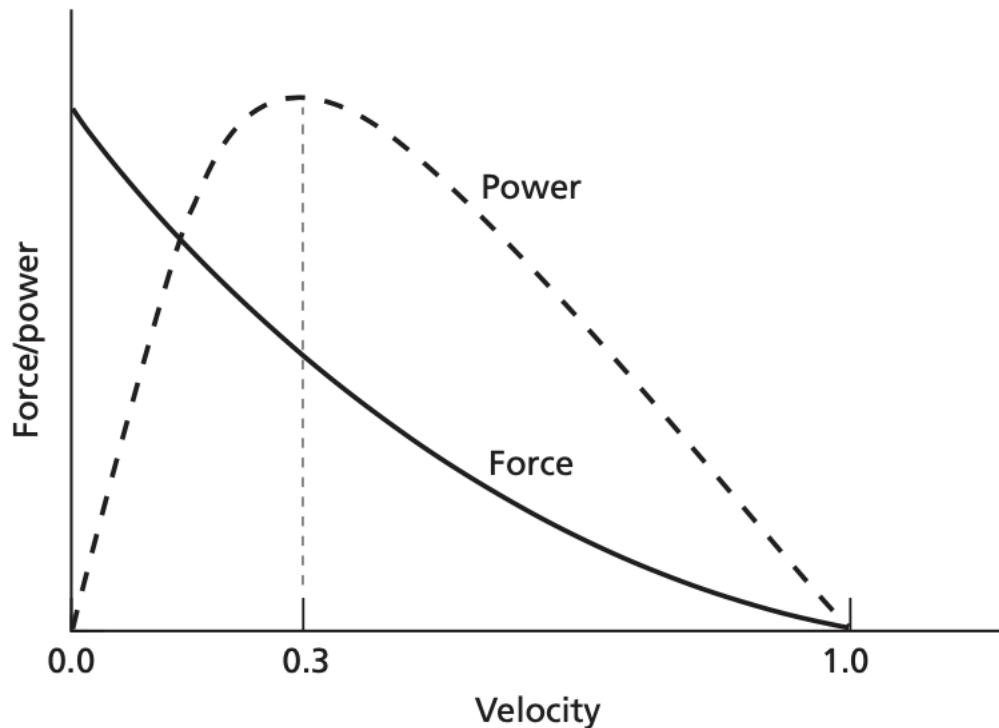
Figure 3: The force-velocity and power-velocity relationship of a muscle. Maximum power is at approximately 30% of maximum shortening velocity. figure from (Herzog, 2000).

Muscle architecture

Both force-length (F-L) and force-velocity (F-V) relationship of the muscle is dependent on muscle architecture. Longer muscle fibers have more sarcomere in series, consequently longer muscle fibers are cable of shortening over greater distances, and the number of sarcomeres in series is proportional to the muscle's maximum shortening velocity. The number of sarcomeres in parallel is proportional to maximal force the muscle can produces, since each parallel sarcomere contributes to force production.

Longitudinal or fusiform have muscle fibers aligned parallel to the line of pull from the tendon. These fibers pull in a straight line and full magnitude of the muscle force is directed in the tendons line of action. In general, muscles that have a parallel fiber arrangement have longer muscle fascicles, and therefore produce a greater range of motion (ROM). Muscles that have a fiber arrangement oblique to the tendon and aponeurosis points of origin and insertion are called pennate muscles. The fibers of pennate muscles attach to the aponeurosis at an oblique angle in relation to the muscle´s line of action (Levangie & Norkin, 2011; Whiting, 2008). The angle of insertion in pennate muscles is named the pennation angle, and

in GM it is usually defined as the angle between the deep aponeurosis and the muscle fascicles (Challis, 2000). This common two-dimensional (2D) definition of pennation angle provides a crude estimation for the fiber orientations in the three-dimensional (3D) muscle.

Pennate muscle fibers can align in one or more directions in relation to the tendon and aponeurosis. Depending on the number of fiber directions relative to tendon and aponeurosis, pennate muscles are defined as either unipennate bipennate or multipennate, the GM investigated in this study is a unipennate muscle (McGinnis, 2013) (Figure 4).

The physiological cross sectional area (PCSA) of a muscle is defined as the line perpendicular to the fascicles line of action (i.e., green line in Figure 4) and is magnitudes lager than the anatomical cross section area (CSA), which is defined as the line perpendicular to the muscles line of action (i.e., blue line in Figure 4). Thus, the main advantage of pennate muscles is that more fibers can be packed into a given volume of muscle, i.e., more sarcomere in parallel. Hence, the total force produced by all the muscles fibers in a pennate muscle is greater than in parallel fibered muscles of the same CSA (Aagaard et al., 2001; V. M. Zatsiorsky & Prilutsky, 2012).

For pennate muscles the force produced by the tendon will be less than the sum of forces generated by the muscle fascicles. Since the force produced by the tendon will be proportional to the force of muscle fascicles multiplied with the pennation angle (α) (Rutherford & Jones, 1992).

$$Force\ tendon\ =\ Force\ muscle\ *\ cos\ \alpha$$

However, given the increased number of fascicles in parallel, Alexander and Vernon (1975) predicted that the tendon force would increase up to an pennation angle of 45° degrees.
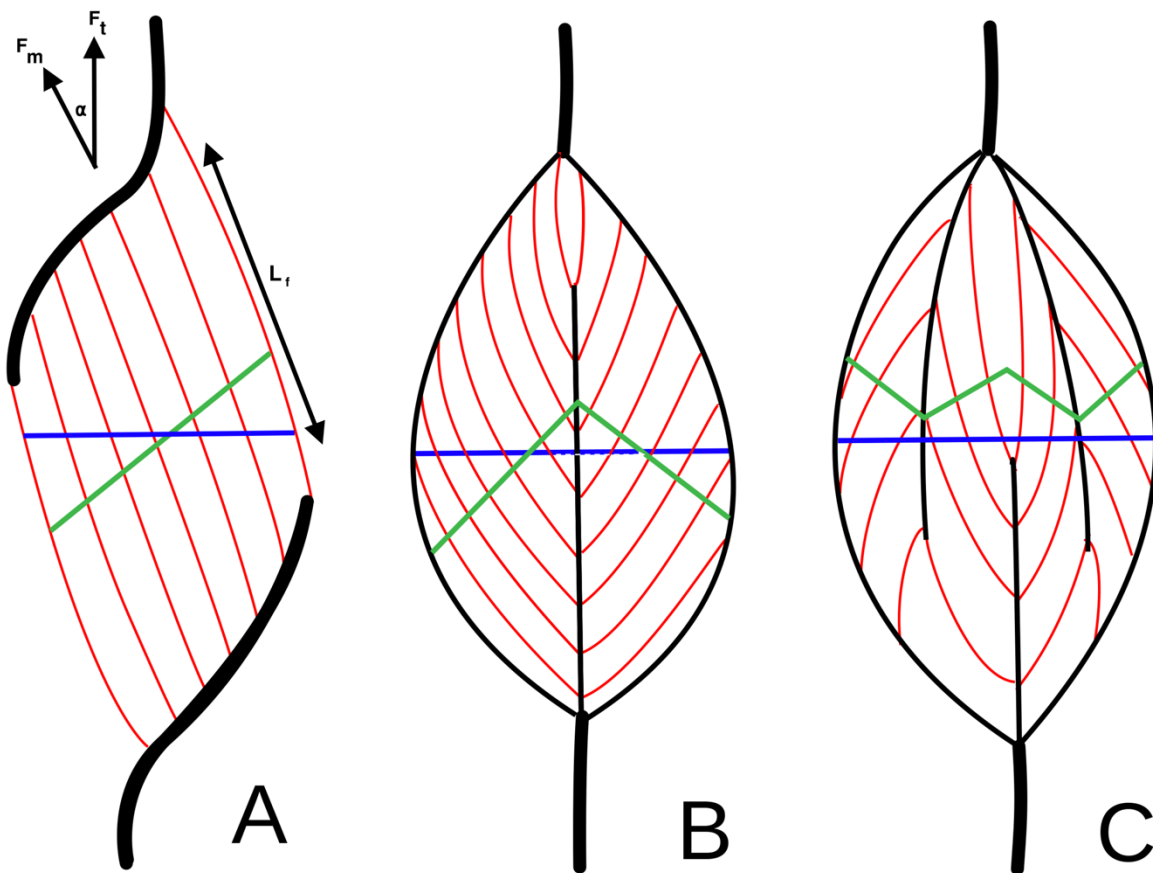
Figure 4: Pennate muscles. A = unipennate., B = bipennate., C = multipennate. Blue line = anatomical cross sectional area (CSA), Green line = physiological cross sectional area (PCSA), L_f = length fascicle, F_t = force tendon, F_m = force fascicles, α = pennation angle. Adopted form Uwe Gille (cc-by-sa)

The Gastrocnemius Medialis (GM) MTU, which is investigated in this study, has its origin at the medial condyle of the femur. The GM, then goes down the medial side of the calf and form a common tendon with the Soleus (SO) muscle, the Achilles tendon (AT) which attaches to the calcaneus. In comparison to the SO muscle, the GM muscle has longer fibers, a smaller pennation angle and a smaller CSA. Thus, the GM muscle can produce lower forces but can operate at higher fascicle shortening velocities and at greater ranges (Challis, 2000).

Tendons/ elastic elements

In pennate muscles, the tendon has an external and an internal part. The internal part of the tendon is referred to as the aponeurosis (V. M. Zatsiorsky & Prilutsky, 2012). The tendons and aponeurosis together are often referred to as the series elastic elements (SEE), however, all elastic elements may not strain in series, since elastic tissue such as the aponeurosis and perimysium may stretch in different direction to the whole tendon. Further, the muscle fascicles are linked and must obey the law of conservation of volume and energy (Gans & Gaunt, 1991), therefore it is more appropriate to refer to the elastic tissues in MTU as the elastic elements (EE). The connective tissues of the muscle weaves into the aponeurosis and tendons that connects muscles to bone. The force of a muscular contraction is thus transmitted from the endomysium of the muscle fibers to the perimysium and epimysium and then to the aponeurosis and tendon. The transition from muscle to tendon is called the myotendinous junctions. The tendon-bone junctions is known as the enthesis (McGinnis, 2013; V. M. Zatsiorsky & Prilutsky, 2012).

Material properties

Like the muscles, tendons have a hierarchical structure (Figure 5). The tendons are composed of collagen molecules, which make up the fibrils, which again make up the fibers, fascicles, fascicle bundles and the lastly the tendon unit. Tendon fibers are bound by the endotenon, and fiber bundles are enclosed by the epitenon. Tendons consist of collagens (mostly type 1), water, elastin, proteoglycans, and glycoproteins. (Wang, 2006).

Collagen molecule

α_1

α_2

Gly X Y

Collagen fibril

64n

280 nm

fibroblast

Collagen fiber

Primary fiber bundle (subfascicle)

Secondary fiber bundle (fascicle)

Endotenon

Tertiary fiber bundle

Tendon Unit

Epitenon

1 nm | 100 nm | 1-20 μm | 20-200 μm | 500 μm
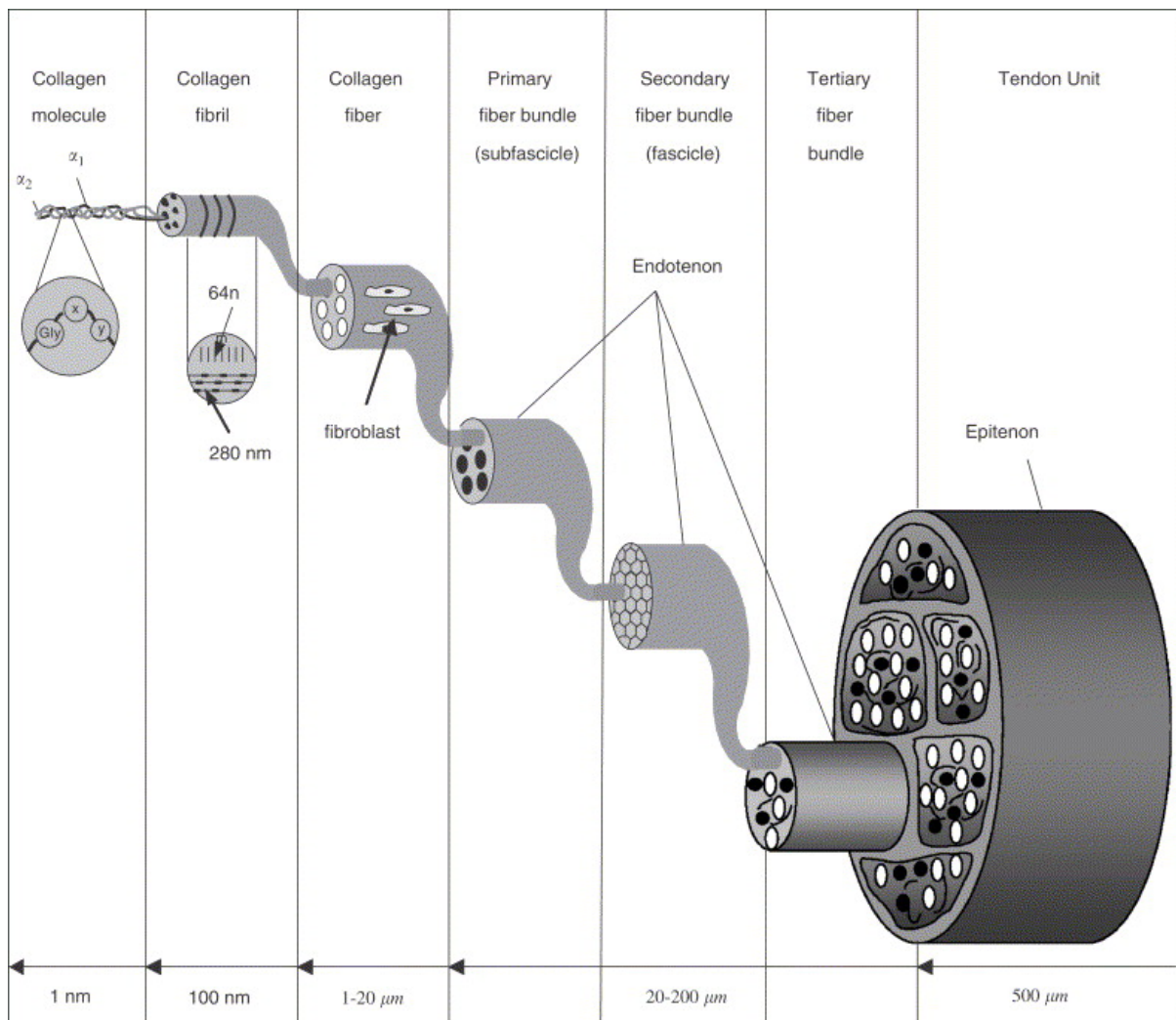
Figure 5 The hierarchical structure of tendons: from (Wang, 2006)

Mechanical properties

The hierarchical structure of tendons aligns collagen fibers in the longitudinal axis of the tendon, giving it its tensile strength. Tendons is a viscoelastic material which when stretch under load is deformed, but return to its original length when force is removed. A tendons elasticity or resilience is its capacity to resist external forces and resume their normal shape when force is removed, i.e. tendon length (V. M. Zatsiorsky & Prilutsky, 2012). Due to its viscoelasticity tendons have another property named hysteresis. Hysteresis is the difference of the loading and unloading phase in the stress-strain cycle. With each loading and unloading phase some energy is lost, mainly by being transformed from mechanical work to heat. Thus, some energy is lost during the storing and releasing of elastic strain energy.

When the tendon fibers are relaxed, they follow a sinusoidal wave pattern known as crimp (Wang, 2006; V. M. Zatsiorsky & Prilutsky, 2012). The tendon stress-strain curve has three regions, the toe-region, the linear region, and the failure region (Figure 6). In the toe-region tendon fibers straighten and lose their crimp pattern, but the fibers themselves are not stretched, this happens in the strain range up to ≈2% strain. After the crimp is straightened, the tendon fibers are stretch, and the tendon enters the linear region on the stress-strain curve. The linear slope of the stress-strain curve is usually referred to as young's modulus (V. M. Zatsiorsky & Prilutsky, 2012).

Stress is the force divided by the area it acts on, measured in newtons per square meter (N/m^2). Strain is the elongations per unit object length. Strain is computed as the ratio of the change in length of a line segment to its initial length (V. M. Zatsiorsky & Prilutsky, 2012).

In the linear region of the stress-strain curve, elongation (Δl) of the tendon has a linear relationship to the force (Δl = cF). where c is proportional coefficient called *compliance* (m/N). However, its often most convenient to use the inverse of compliance *stiffness* (N/m) (V. M. Zatsiorsky & Prilutsky, 2012). Compliance or stiffness is measured in the linear region of the stress-strain curve, as seen in Figure 6.

The more a tendon is strained the more elastic energy it can potentially store, thus compliant tendons have a greater capacity for storage and recoil of elastic energy. A stiffer tendon, however, can transfer more of the force produced from the muscles into joint movements. Ettema (2001), showed that muscles with stiff tendons are most efficient when activated during shortening. Whereas, a compliant tendons is most effective when the muscle is activated during the stretch in a stretch-shortening cycle.

The Achilles tendon, which is the shared tendon of the Gastrocnemii muscles and the soleus muscle is incredibly strong with reported froce values reaching 9 KN, corresponding to 12.5 times body mass (Komi, 1990). Further, Biewener and Roberts (2000), and Farris and Sawicki (2012a) has described the Achilles as long and compliant tendon well suited to store and recoil elastic strain energy.
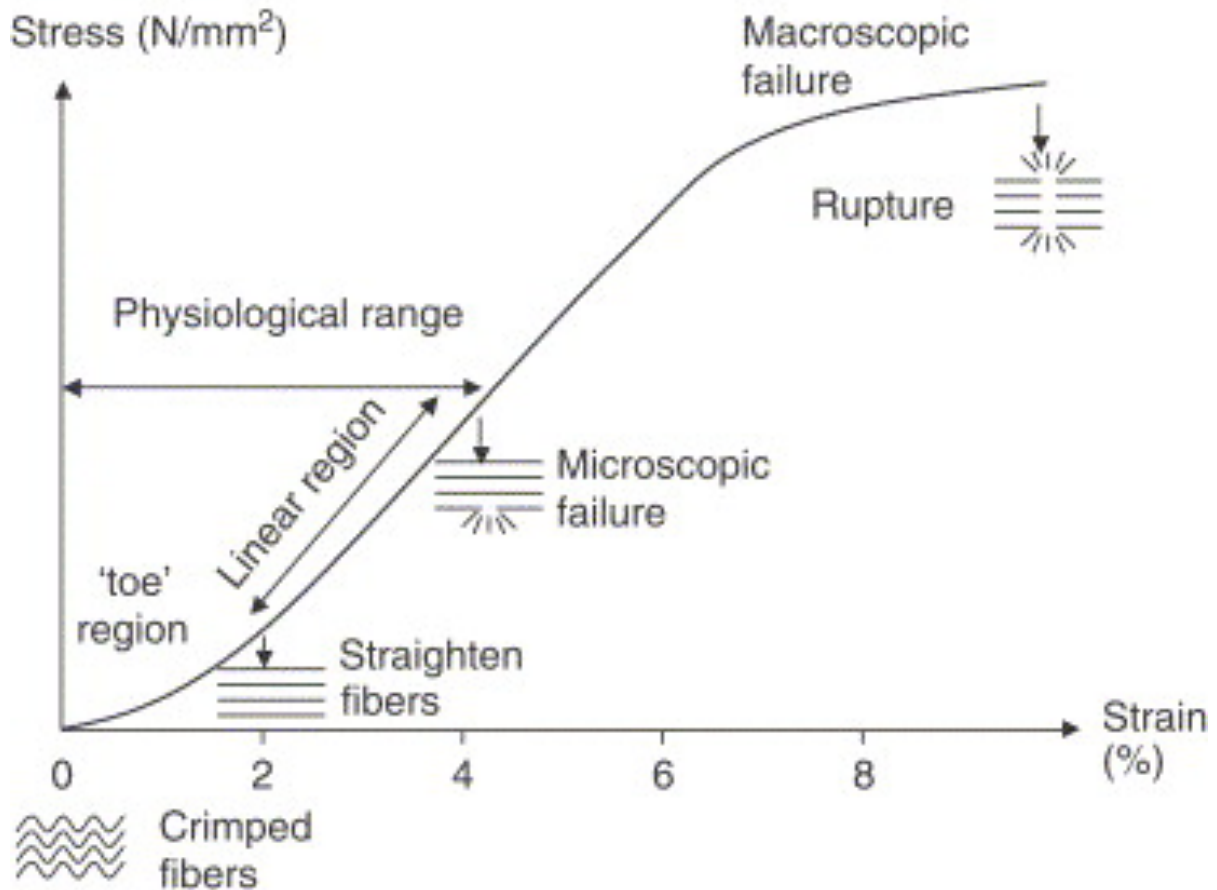
Figure 6 Stress-Strain Curve (Wang, 2006)

Muscle-tendon Unit

A muscle together with the attached tendon is called the muscle-tendon unit (MTU).
The plantar flexor MTU investigated in this study has previously been described as having an architecture with relatively short and highly pinnate fascicles and long compliant tendons, well suited to generate force economically (Biewener & Roberts, 2000; Farris & Sawicki, 2012b).

The muscles potential to generate force depends on the distinct properties as the F-L and F-V relationship. Compliant tendon and aponeurosis enable fascicle length changes to be decoupled from that of the whole MTU. The decoupling of muscle fascicle length from that of the whole MTU enables muscle fascicles to work nearly isometrically and close to their optimal length, regardless of the significant length changes of the whole MTU, thus the muscle fascicles can generate higher forces economically, and at lower metabolic cost (Lichtwark et al., 2007). An equally important part of decoupling to the MTU´s capacity to

generate force is the difference in distinct properties between the CE and the EE. Whereas the muscle fascicles are limited by the rate of the enzymatic processes associated with cross-bridge cycling, the EE mechanical function is based on its structural basis of collagenous fibers, and therefore it does not have the same power and shortening velocity limitations as the muscle fascicles. Thus, muscle work applied to EE in a relative slow contraction, can be rapidly released, and vastly exceeding the power and shortening velocities of the muscle fascicles (Roberts & Azizi, 2011).

In their commentary article, Roberts & Azizi (2011), describe the roles biological springs have in vertebra movement (Figure 7). Or in other words, how the EE of MTU has a diverse set of roles in locomotion. When the EE is stretched under load, it can store elastic strain energy, which later in the stance phase, when the EE recoils, can be released. Further, Roberts & Azizi (2011) describe that the most prominent benefit of springy gait (i.e. utilization of elastic strain energy) is that the work done by the tendons does not have to be done by the muscles, and that muscle work is metabolically expensive. Studies on running turkeys have found that the work done by the EE can be almost twice that of the CE, thus tendons effectively decrease the work needed by muscles fascicles (Biewener & Roberts, 2000). However, tendons can only store energy when the muscles fascicles generate force, thus there is a cost to saving and recoiling elastic energy (Roberts & Azizi, 2011). The Fenn effect, states that active muscles use more energy when performing work, than when only generating force (Fenn, 1924). Since tendons allow muscle fascicles length to be decoupled from that of the whole MTU length, muscle fascicles operate almost isometrically and at a favorable force-length-velocity (F-L-V) relationship (Farris & Sawicki, 2012a; Lichtwark & Wilson, 2006). Thus, tendon mechanics allow the muscle fascicle to generate more force, and by that lowering the required active muscle volume to create sufficient support forces (Roberts & Azizi, 2011). This is in line with the findings of Bohm & colleagues (Bohm et al., 2019) who found an inverse relationship between energetic cost of human running and the F-V relationship in the SO muscle. Giving evidence for that the fascicle shortening velocity of the soleus is a determining factor of running economy.
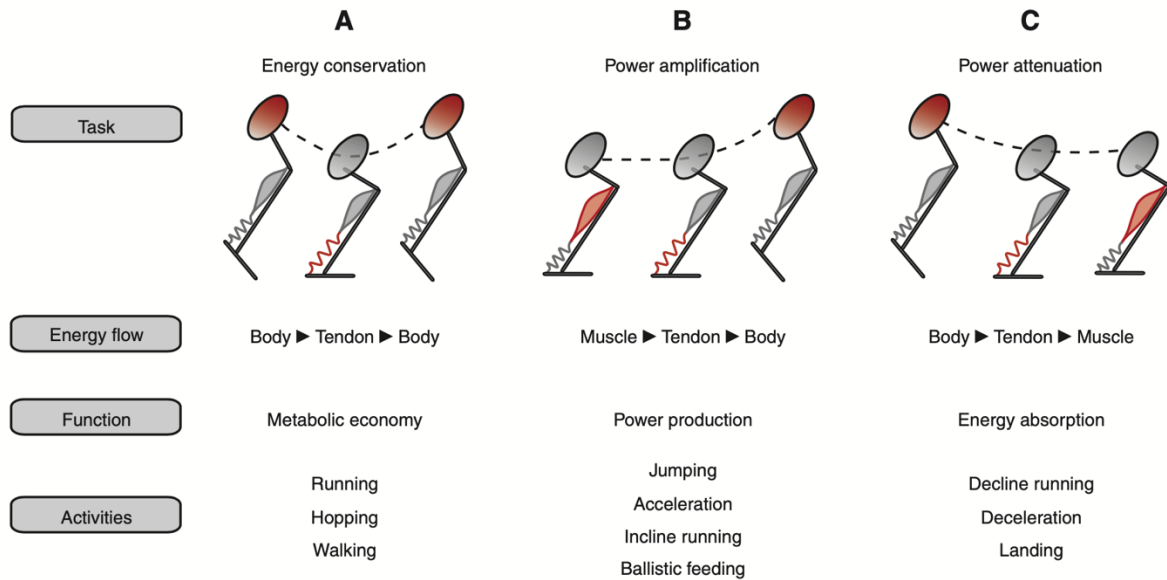
Figure 7: illustration of the triceps surae during different task: (A) Energy conservation, (B) Power amplification, (C) Power attenuation. Figure from (Roberts & Azizi, 2011).

Several studies have showed the elastic energy conservation occur in the triceps surae during walking (Cronin et al., 2013; Fukunaga et al., 2001), running (A. K. M. Lai et al., 2018; Lichtwark et al., 2007; Monte et al., 2020) and hopping (Lichtwark & Wilson, 2005). These studies showed that; at ground contact, muscle fascicles shortened at relatively low velocities thus acting almost isometrically. At the same time the EE lengthened in the first part of the stance phase, before subsequently recoiling in the later part, and thus returning most of the energy stored in the tendons during lengthening. Since tendons doesn´t utilize chemical energy in form of ATP, the contributions of elastic energy from the EE increase running efficiency.

Tendons loaded directly by the contraction of muscle fascicles can release the energy rapidly to the body. If the energy is released more rapidly than it is stored, muscle power can be amplified (Roberts & Azizi, 2011). The energy released by the tendon is equal to, or slightly less (given some energy loss/ hysteresis) the work done by the muscle, however it can be released in a shorter amount of time, thus increasing power output (power = work/time). This mechanism makes it possible for the MTU to produce power outputs vastly exceeding the maximal power output of the muscle fascicles.

Muscle-tendon unit interaction in locomotion

Lai et al., (2014, 2015), found that compliant tendons contributes the bulk of the length change in the MTU, thereby effectively reducing the muscle fascicle shortening velocity and allowing a more favorable F-L-V curve to be maintained during the propulsion phase of running. Further, they showed that the minimal shortening of the fascicles compared to that of the whole MTU was a factor irrespective of running speed. This is in line with Bohm et al., (2019) who found that the decoupling of fascicle trajectories from that of the MTU resulted in a 4.5-fold reduction in the fascicle operating velocity during running. During one legged hopping Lichtwark and Wilson (2005) found that the AT was able to return approximately 76% of the energy stored in the downward motion and recoil this in the upward motion. Thus, providing substantial amount of the total mechanical energy in the hop (16%). Typical length change of the gastrocnemius MTU and fascicle lengths during on legged hopping is showed in Figure 8. here we can clearly see that the tendon does the bulk of the length change in the MTU.
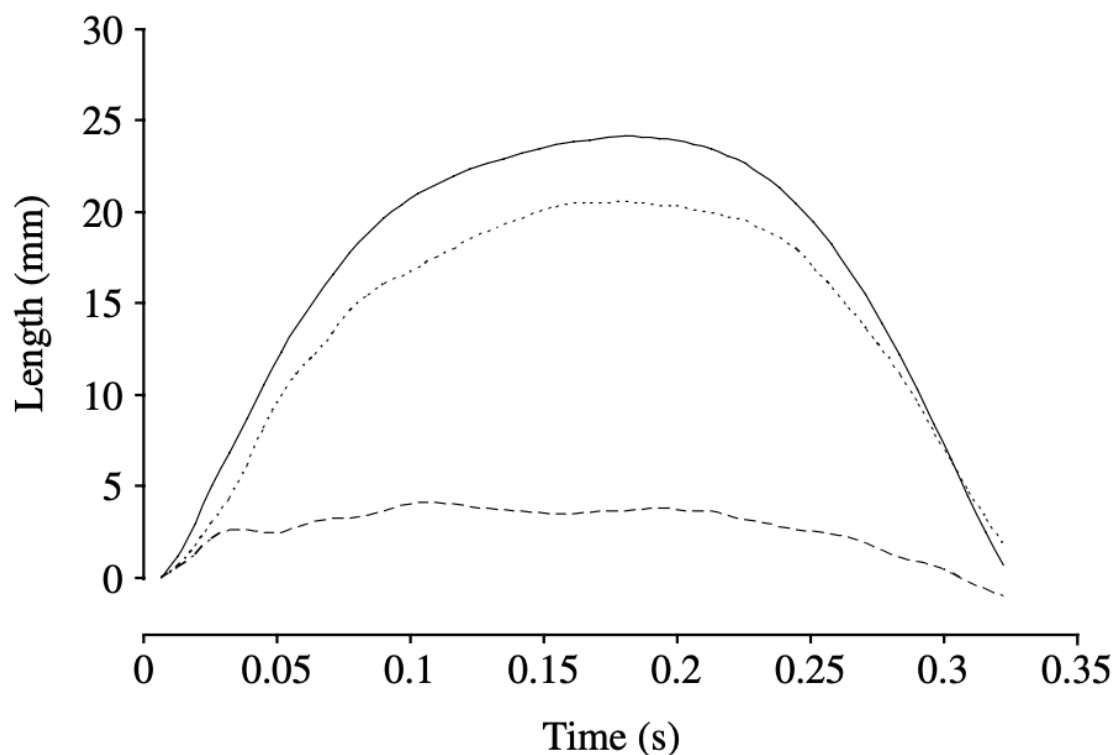

Figure 8 This figure shows the average change in Achilles tendon length (dotted line). The gastrocnemius muscle-tendon unit length (solid line), and the approximate fascicle length during one legged hopping. From (Lichtwark & Wilson, 2005)

Increasing speeds

Dorn et al., (2012) and Lai et al., (2015) found that when running speed increases, so does the muscle fascicle shortening velocity. However, the muscle fascicle shortening velocity was relatively low compared to that of the whole MTU, with the rate of shortening for the fascicles never surpassing 30% of that of the MTU. And the maximum shortening velocity was always less than 3 lengths per second ($L_0 \ s^{-1}$), which would be less than 1/3 of the assumed maximal shortening velocity at approximate 10-13 $L_0 \ s^{-1}$ (Zajac, 1989). Lichtwark et al., (2007) found that during running at 7.5 km h^{-1}, the muscle fascicles shortening velocity was 0.54 $L_0 \ s^{-1}$. Considering the theoretical maximal shortening velocity described previously this is very slow, and thus the muscle fascicles operate at an optimal level for high force with small amount of work and high efficiency (i.e., favorable position on the F-L-V curve). While the muscle fascicles operated at slow speeds, the whole MTU, due to the stretch and recoil of the tendon operated at approximately 6 muscle fascicle $L_0 \ s^{-1}$. Similarly, Lai et al. (2015) found that fascicle shortening velocity increased continuously when running speed increased from 2.0 – 5.0 m s^{-1}. Whereas the MTU shortening velocity increased up to a speed of 4.0 m s^{-1}, but remained unchanged from 4.0 m s^{-1} to 5.0 m s^{-1}.

Increasing incline

Lichtwark and Wilson (2006), found an increase in muscle fascicle length with increase in grade. This was likely due to the increased in the whole MTU length occurring because of a more dorsi-flexed ankle, as well as changes in the force requirements. Thus, the muscle fascicle length during force production increased.

At level running, Farris and Sawicki (2012a), found that the ankle accounts for 42-47% of the net positive power. In line with this, Roberts and Belliveau (2005) found that at level running the hip contributed almost nothing to the net power, thus the power came mainly from the knee and ankle plantar flexor. However, at 12° incline the hip performed ≈75% of the total net work because of its increased moment arm. This means may suggest that as incline increases, the role of the plantar flexors becomes less important

Biomechanics of diagonal stride XC-skiing

The use of skies allows for gliding/rolling in a substantial part of the gait cycle, accounting for about 47% of cycle duration at 2° incline and 37% of cycle duration at 10° incline (Pellegrini et al., 2014, 2020), which increase stride length. Propulsion is done by both legs and arms through the use of skies and poles, thus XC-skiing can be viewed as a four limbs gait (Pellegrini et al., 2014). In diagonal stride the arms and legs move in a coordinated pattern that resembles running. During each cycle there is a substantial glide phase, , this is followed by a propulsive action of the leg, described as a backwards kick when the skies motion in relation to the ground surface is zero, this requires a certain amount of static friction (Pellegrini et al., 2020). When skiing on snow this friction is created by pushing down the ski camber so that the wax can "grip" the snow. Thus, grip when skiing on snow varies with snow conditions, wax type, stiffness of the ski camber as well as the skiers technique. However, roller skies have a ratchet wheel mechanism, which only allows the wheel to move in the forward direction, giving roller skies almost unlimited grip.

Energy dissipation

Kehler et al., (2014) found that DIA initially appear similar to running with an in-phase energy fluctuations, but in contrast to running, Kinetic energy (KE) is lost to friction in the glide phase. Work lost to the environment due to drag or friction cannot be stored and recovered within the EE to help power cyclical movements (Roberts & Azizi, 2011). Energy lost in skiing is primary due to ski/snow friction or roller resistance and aerodynamic drag force. In their commentary article, Roberts and Azizi (2011) notes that during locomotor movements, the storage of muscle work in elastic elements may allow muscles to perform work during a period e.g., the glide phase, when the application of power to the body is constrained by kinematics or kinetics. Accordingly, it may be possible to store muscle force in the tendons during the glide phase, even though the mechanical energy is dissipated.

Muscle mechanics during diagonal stride XC-skiing

In a review article, Zoppirolli et al., (2020) found that cycle length is related to better skiing performance and skiing economy in DIA. Further, maximal speed was associated with greater propulsion forces by the legs as well as shorter relative phase of leg propulsion. Thus, increased power due to favorable muscle-tendon dynamic may increase performance

Komi and Norman (1987) theorized that there is a similar muscle-tendon dynamic in DIA, as seen in running. They reported a stretch-shortening cycle of the MTU during DIA. Further, Kehler et al.,(2014) theorized that a preload phase prior to the kick may facilitate elastic energy storage, that may contribute to a more beneficial muscle-tendon dynamic, by conserving some of the gravitational potential energy (GPE) and amplify the force output in the late part of kick.

Studies on muscle-tendon behavior in DIA XC- skiing measuring the fascicle length and shortening velocity directly has not been conducted yet. Advances in technology the recent decades have seen new and more accurate ways to examine muscle-tendon unit behavior. With ultrasound imaging we can directly measure the muscle fascicles length and length changes in vivo during gait. Using three-dimensional (3D) motion capture we can estimate the 3D kinematic changes of the MTU. Therefore, it is now possible to accurately study the muscle-tendon behavior and whether it store and recoil elastic energy.

Materials and methods

Participants

13 elite senior XC-skiers (8 male and 5 female) participated in this study. Table 1 show the participants characteristics. The participants were required to have at least 5 years of experience in XC-skiing and roller skiing. Further the participants were required to compete or recently have competed at least on the level of either Norwegian XC Cup, the Scandinavian XC Cup or the World Cup. All participants were familiar with roller skiing on a treadmill. Their level varied from world class (n=7) to skiers at a high national level (n=6).

Table 1. Participants anthropometrics

	Mean	$\pm$ SD
Age (years)	26.5	4.8
Height (cm)	179.3	8.0
Body mass (kg)	72.8	6.2

Ethics statement

Prior to the study an ethical approval from The Ethics Committee of the Norwegian School of Sport Sciences was obtained (case number: 22102017). The approval for collection and storage of data for this project and it´s security, was obtained from the Norwegian Center for Research Data (NSD) (case number: 571816). All the participants were informed about the experimental protocol and potential risk of the study, as well the right to withdraw from the study at any point without providing a reason, through an information letter. All participants provided written consent. Therefore, the ethic of this study was done in accordance with the Helsinki-declaration and the ethical requirements of the institutions.

For ethical approval and information letter see appendix 1

Experimental protocol

This was an experimental test, done under controlled conditions in a biomechanics lab. All test on a participant was done on one day, and in total they spent around one and a half hour in the lab.

Muscle-tendon mechanics, kinematic, kinetic, and muscle activity measurements were performed while the participants roller skied on a motorized treadmill (size 4.5 m x 3 m) (Rodby, Vänge, Sweden) using DIA under three different conditions. All the participants used the same pair of roller skies (IDT Sport classic, Lena, Norway), the ratchet wheel mechanics was moved from the back wheel to the front wheel, since this would (at least in theory) require the skier to push more perpendicular to the surface, thus creating a more "on snow" like biomechanics. Participants used poles with ferrules (spikes) designed for treadmill use. The participants self-selected the pole length and pole length was in accordance with the international regulations (83% of height) (FIS, 2020).

The participants followed a standardized warm-up of 10 minutes, (7 min at 2.5ms^{-1} and incline of 5°, one min at 10°, then 30 second at 3.5ms^{-1} at 5°, then back to 2. 5ms^{-1} for one min). The warm-up included at least 30 seconds on each of the conditions such that the participants could get used to the speed and incline. Since roller skies open up to different biomechanical solutions than on snow skiing, the participants were asked to mimic the on-snow technique to the best of their ability.

The test protocol included three conditions with two trials per conditions:

- The first condition was diagonal stride at an incline of 5°, and a relative low speed at 2.5ms^{-1}. Hereafter called DIA_{ref}
- The second condition was diagonal stride at an incline of 5°, and a relative high speed at 3.5ms^{-1}. Hereafter called DIA_{fast}
- The third condition diagonal stride at an incline of 10°, and a speed of 2.5ms^{-1}. Hereafter called DIA_{steep}

For each condition the participant accelerated up to desired speed, then data was collected for 10 seconds. The data collection started with a trigger signal from the ultrasound (US) measurement making sure that the kinematic, Kinetic and EMG data were synchronized to US data. After the data collection the participant was deaccelerated to a full stop. Between each trail there was a 2-minute break where data was saved, and the data programs reset (Qualisys Track Manager (QTM)). Thus, giving the participants enough time to rest between trials Figure 9 shows the warm-up and test protocol.
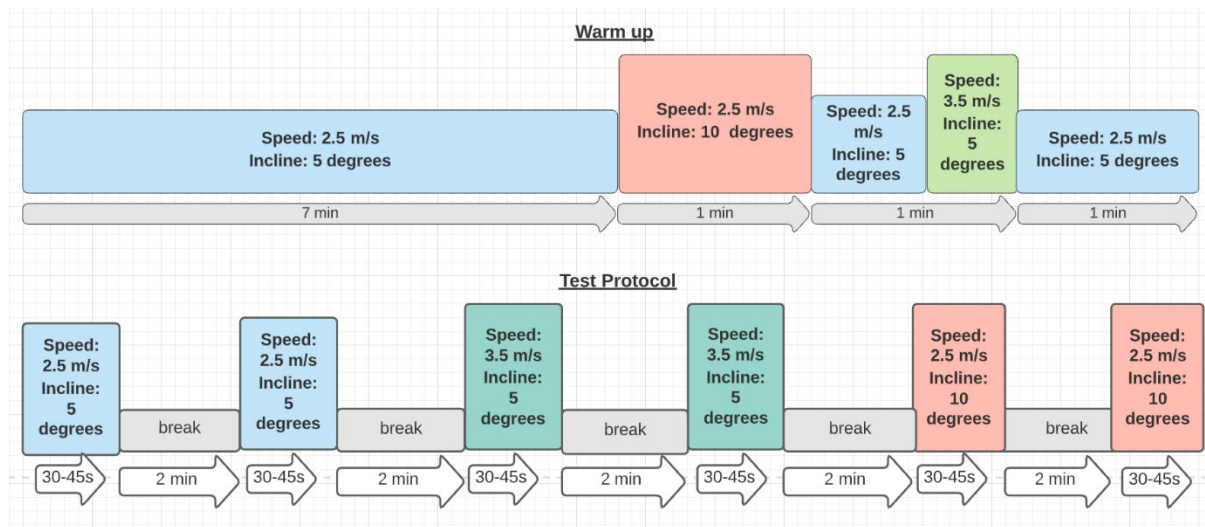
Figure 9: Warm up and test protocol

Measurement of muscle mechanics

Ultrasound

The GM muscle fascicles were imaged using an ultrasound linear array transducer (ArtUs, Telemed, Vilnius, Lithuania). The transducer was placed to the leg using a custom-made holder with self-adhesive bandage wrapped around the leg. B-mode images with a field of view of 50 mm and depth of 65cmm were collected at 117 Hz.

To measure the fascicle length and pennation angle the transducer was placed over the GM muscle mid belly and aligned with the aponeurosis

For the cables of the US measurement system to not interfere with the poles during trials the US control units were placed in a custom-made plastic box with Styrofoam. The box was placed above the participants by an adjustable pulley system. The cables from the US control unit went down the participants back and were secured around the participants abdomen by and elastic band (Figure 10).

Figure 10. Experimental setup: Participant on the treadmill with ultrasound-transducers, reflective markers, electromyography (EMG) transmitters and force in ski measurement attached. The participant wore an ultrasound-transducer on the gastrocnemius medialis GM and vastus lateralis (VL), however only data from the GM is presented in this study. The cables were held in place by elastic bands around the ankle, knee and abdomen, and ultrasound and force control unit was placed in a box above the participants.

Fascicle length and pennation angle

A semi-automated tracking algorithm (UltraTrack_v5_2_2.m) was used to analyze fascicle length and pennation angle (Farris & Lichtwark, 2016)(Cronin et al., 2011). The fascicle and aponeurosis were clear and visible in all subjects. However, a frame-by-frame analysis with manual correction of fascicle and aponeurosis whenever the tracking algorithm was insufficient was performed. Manual corrections were performed by the same researcher for all participants.

Fascicle length was defined as being the length between the fascicle insertion of the superficial aponeurosis to the deep aponeurosis, under the assumption that the fascicle trajectory is linear. Linear extrapolation was used to define fascicle length when fascicle insertion exceeded the ultrasound imaged. Pennation angle was defined as the mean of the angle of insertion between the fascicle and the deep and superficial aponeurosis. (Figure 11)
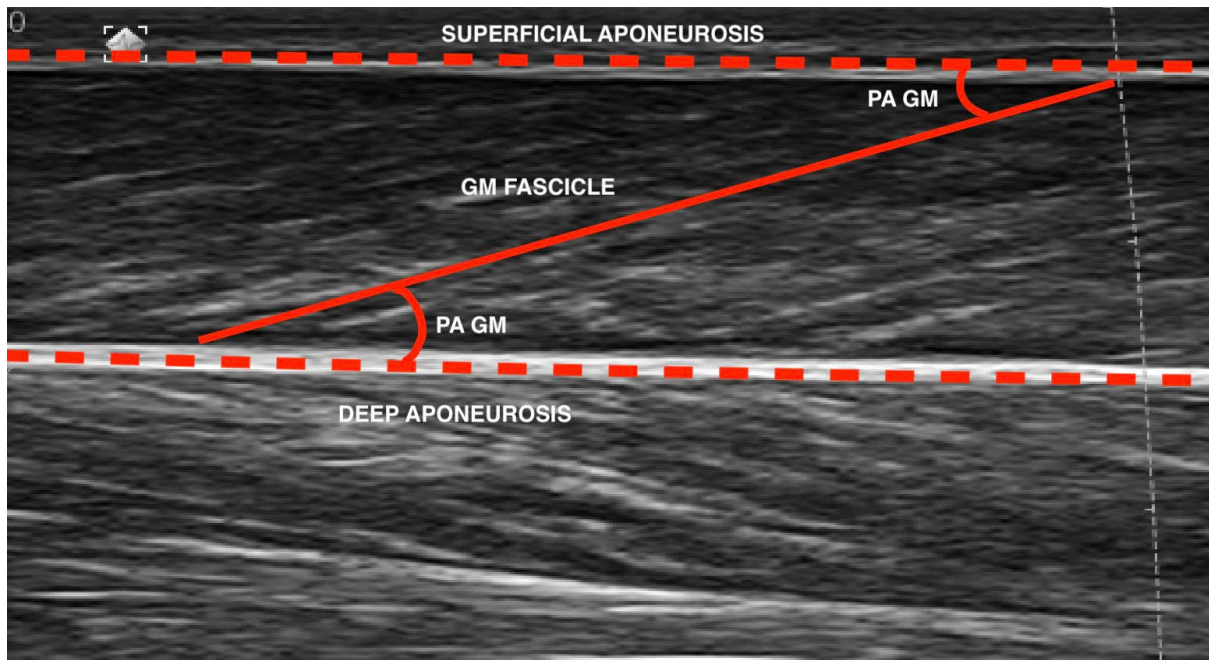
Figure 11: Ultrasound imaging of the gastrocnemius (GM) muscle and aponeurosis. PA GM = Pennation angle gastrocnemius medialis. The pennation angle used in this study is calculated from the mean angle of the fascicle insertion on the deep and superficial aponeurosis.

Muscle fascicle lengths measurements may vary depending on where you put the ultrasound probe (Lai et al., 2015; Lichtwark et al., 2007). Lichtwark et al (2007) showed that the distal fascicles tend to shorten more and work at greater pennation angles than the more proximal fascicles. Thus, the muscle fascicle length is not homogenous throughout the muscle. However, the authors concluded that the mid-belly of the muscle provides a good estimation of the average length changes across the muscle. Thus, this is where the probe was placed in this study.

MTU length

MTU GM length was estimated using the formula form (Hawkins & Hull, 1990), which is a model for estimating lengths of lower extremity muscle-tendons units during flexion/ extension movements:

$$L = C0 + C1\chi + C2\beta + C3\beta^2 + C4\phi.$$

In this expression L represent the normalized muscle-tendon length, C0 to C4 are regression coefficients, and $\chi, \beta \ and \ \phi$ are the hip, knee, and ankle flexion angles respectively in degrees.

For the formula for gastrocnemius medialis is as follows:

$$MTU\ GM\ =\ 0.900\ -\ 6.20E-4\ *\ (knee\ angle)\ +\ 2.14E-3\ *\ (ankel\ angle\ +\ 90).$$

Ankle, knee and hip angles was calculated in **Visual 3D** (C-motion, Germantown, MD, USA) The GM Fascicle shortening velocity (dFL) and MTU GM shortening velocity (dMTU) was calculated from numerical differential of FL and MTU, respectively.

EE length

To estimate the length of EE, length of the muscle fascicles in the direction of the tendon was subtracted from the entire MTU length (Fukunaga et al., 2001; Lichtwark et al., 2007):

$$EE\ GM\ =\ MTU\ GM\ -\ (Fascicle\ length\ *\ cos(pennation\ angle\ *\ (\pi/180)).$$

This model, used in most research to calculate the EE length, assumes that all EE work in series with the muscle fascicles (Fukunaga et al., 2001).

Shank length

Shank length was defined as the mean distance between the Right lateral femoral epicondyle (RLEPI) and the Right lateral malleolus (RLM) using the formula for distance between two points in 3D space:

$$d\ =\ ((x_2\ -\ x_1)^2\ +\ (y_2\ -\ y_1)^2\ +\ (z_2\ -\ z_1)^2)^{1/2}.$$

Kinematics

Kinematic data was collected at 150Hz using the Qualisys system with 12 infrared Opus 400 and 700 cameras (Qualisys AB, Gothenburg, Sweden). The 12 cameras mounted at different heights and angles encircled the treadmill to make sure all markers in the area of movement was registered.

Spherical reflection markers (12mm, Qualisys AB) were placed in accordance with (Robertson et al., 2014) at key anatomical position at the hip and right leg, and additional reflective markers where placed on relevant sites of the equipment. Complete marker list can be found Table 2. The foot markers were placed on the outside of the shoe. Reflective details on shoes and clothing were masked with tape as to not interfere with the other markers.

Tracking markers (cluster of three reflective markers on a custom-made triangular base) were placed on the thigh and shank of the participants with two-sided tape as well as sport tape.

Focus and aperture were adjusted for each camera prior to the first test. The system was calibrated dynamically using a calibrations wand (750 mm) and a stationary calibration frame consisted of 4 markers in an L shape (X, Y, Z), during a calibration period of 45 seconds. In this period the wand was constantly moving within the estimated volume of movement in the trials . The calibration was successful when a sufficient volume was covered, and the estimated average residual value was below 1mm (computed by the Qualisys Track Manager (QTM) system). Average residual value was 0.511 ± 0.041 mm.

The coordinate system was as follows: Y was defined as the same direction as the locomotion and positive in forward direction, Z-axis was perpendicular to the treadmill belt with positive direction pointing upwards. The X-axis was horizontal with positive direction being right. With changing incline, we used a function in MatLab (version 9.9.0 (R2020b), The MathWorks Inc., Natick, MA, USA) to rotate the coordinate frame so that the y-axis is parallel to the locomotion forward direction and the z-axis is perpendicular to the treadmill band. The x-axis remains horizontal.

Markers were identified using an automatic identification marker (AIM) model, gap fill function was used when needed. In one static trial, used to create the skeletal model in visual 3D, one marker was missing (RMEPI) and a special script in MatLab was used to gap fill this marker.

Table 2. Description of the reflective marker placement

Segment	Marker	Position
Body markers	RASIS	Right anterior-superior iliac spine
	LASIS	Left anterior-superior iliac spine
	RPSIS	Right posterior-superior iliac spine
	LPSIS	Left posterior-superior iliac spine
	RLEPI	Right lateral femoral epicondyle
	RMEPI	Right medial femoral epicondyle
	RLM	Right lateral malleolus
	RMM	Right medial malleolus
	RHEEL	Right heel (i.e., most prominent point)
	RTOE5	Right fifth metatarsal head
	RTOE1	Right second metatarsal head
	RTOE2	Right first metatarsal head
Segment markers	RTH 1-3	Right thigh segment marker
	RSH 1-3	Right shank segment marker
Equipment markers	RSA	Right roller ski anterior
	RSP	Right roller ski posterior
	LRSA	Left roller ski anterior
	LRSP	Left roller ski posterior
	PRU	Pole right upper (15 cm below handle)
	PRL	Pole right lower (60 cm above spike
	PLU	Pole left upper (15 cm below handle
	PLL	Pole left lower (60 cm above spike)
	TPA	Treadmill position anterior
	TPP	Treadmill position posterior

Muscle activity

EMG data was recorded using a telemetered system (Aktos, Myon, Schwarzenberg, Switzerland) with a sampling frequency of 2100 Hz. Bipolar surface electrodes (24 mm diameter) (Kendall Arbo H124SG electrode, Coviden, Minneapolis, USA) were mounted on the skin overlying the muscle belly of Vastus lateralis (VL), Biceps femoris (BF), Gastrocnemius lateralis (GL), Soleus (SO) and Tibialis anterior (TA) in accordance with Seniam (Surface Electromyography for the Non-Invasive Assessment of Muscles) recommendations (Hermens et al., 2000). Prior to the fastening of the EMG electrodes the skin was shaved using a one-time razor and rinsed with isopropanol. The EMG transmitters (Aktos waterproof EMG transmitter, Myon Schwarzenberg, Switzerland) were connected to the electrodes and secured using double-sided tape and sports tape.

To check the EMG signal for clarity and strength the participants were asked to perform isolated dynamic muscle movements. The participants were asked to perform a squat (VL), a leg curl (BF), a calf raise (GL and SO) and an ankle dorsi flexion (TA)

EMG was normalized to the maximum processed EMG value (EMG_{MAX}) obtained from the participants during DIA_{ref}. Threshold value for muscle activity was set to 25% of peak based on the recommendations of *Özgünen et al.* (2010).

Force measurements

To get force measurements from the ski, we put two Force Sensitive Resistors (FSRs, Flexiforce A201, Tekscan) under the ski binding measuring at 2100Hz and load range 0-445N. To ensure that all forces between foot and ski were transmitted through the FSRs, we built a custom-made rectangular disc with a circular piston (10mm) to press against the FSR's sensing areas. We had one FSR placed under the forefoot, and one placed at the heel.

The FSR was connected with the Telemyo 2400T G2 transmitter (Noraxon, Arizona, USA) via an extra-long cable (same cable management as US). The FSR was secured with double-sided tape and the cables were secured to the ski with plastic strips and to the leg of the participant and around the participants back with elastic bands.

Between warm-up and trials, three calibration measurements of the force in ski were conducted. One zero force measurement where the participant lifted the right ski of the ground. One heel measurement where the participant balanced his/her bodyweight on the heel. And one forefoot measurement where the participant balanced his/her bodyweight on the forefoot. Offset for each sensor (b_1 and b_2) was acquired from the first recording (ski off ground). The two gains (s_1 and s_2) were determined by solving the system of equations:

$$1 = s_1(x_{1,1} - b_1) + s_2(x_{2,1} - b_2),$$
$$1 = s_1(x_{1,2} - b_1) + s_2(x_{2,2} - b_2),$$

where $x_{i,j}$ and $x_{i,j}$ are the FSR outputs from sensor i (forefoot or heel) and recording j (weight on forefoot or weight on heel). Hence, the sum of the two FSRs after calibration should equal force normalized to body weight.

Data processing and statistics

Phase division

We divided one cycle (i.e., from initiation of kick to next initiation of kick) into three phases. The kick phase, where forward propulsion is generated, the swing phase where the leg is relocated, and the glide phase where participants glide/ roll on the skies. Furthermore, we divided glide into two equal phases, glide first and glide last.

Kick phase

Kick phase was defined as when the ski speed (v ski) was equal to that of the treadmill speed (i.e., the ski is not moving relative to the ground surface). In accordance with Pellegrini et al., (2014, 2020) ski speed was defined as equal to the treadmill speed when v ski remained below a threshold of 0.5 km h^{-1} relative to the treadmill belt. We used the reflective marker placed anterior on the right roller ski (RSA) as reference for the position of the ski in Y-axis. Then, we used the derivate of position data of the roller ski marker, thus giving us the velocity in m s^{-1}. This multiplied with 3.6 gives us the speed in km h^{-1}. The velocity was put through a low-pass Butterworth filter with a cut-off frequency of 20 Hz

Glide phase and swing phase

The start of swing phase was defined as end of kick (i.e., v ski > 0.5 km h^{-1}), whereas the start of glide phase was defined as the greatest force increase form the force ski measurement,

calculated from the numerical differential of the force curve. Further, glide phase was divided in the first half and second half of the glide phase, referred to as "glide first" and "glide last".

Time normalization

All data was time normalized to 101 points per cycle. The start of the cycle was defined as initiation of the kick, with the next cycle starting at the initiation of the subsequential kick. Then the average of five complete cycles for all variables were used further for the analyses.

Filter and Processing

Filtering of kinematic and ultrasound data was performed with a lowpass digital zero-phase Butterworth filter with a cutoff frequency of 15 Hz. For force filtering, a cutoff of 30 Hz was used. EMG data was filtered through a bandpass Butterworth filter with low cutoff of 10Hz and high cutoff of 500Hz, then moving root mean square (RMS) was calculated. Linear interpolation was performed on ultrasound data resampling it to the same frequency as the kinematic data.

Statistical analysis

Statistical test was performed using Prism (Version 9.1.1, GraphPad Software, LLC, San Diego, CA, USA)

To test our first hypothesis, we used a single-sided One Sample T-test. All variables were check for normality using a Shapiro-Wilk's test ($p > .05$). Whenever variables were not normally distributed a Wilcoxon signed-rank test was performed. The hypothetical value was set to 0 for all variables except for force and EMG, which was set to 1 for force, and 0.25 (25% of peak) (Özgünen et al., 2010) for EMG. Statistical significance was set to (P<.05), and data are presented as mean ± standard deviation (SD).

For our second hypothesis, a repeated measures One-Way ANOVA with a Geisser-Greenhouse correction was performed to test the effect of increased speed and incline. We then performed a Šídák post hoc test where we compared the mean of each column with the mean of DIA_{ref}. Statistical significance was set to (P<.05), and data are presented as mean ± SD.

Statistical power

Statistical power analysis was conducted prior to the study using G*Power (Version 3.1.9.6) (Faul et al., 2007)(Faul et al., 2009) to estimate a sufficient sample size, and to estimate effect size. US Fascicle data from running at different speeds and incline was used as an estimate for expected values (mean $\pm$ SD) during diagonal stride at different speeds/ incline since there is currently no available fascicle data in XC-skiing. The data was retrieved from (Bohm et al., 2018; A. Lai et al., 2015; Lichtwark & Wilson, 2006) and consisted of GM fascicle length and maximum shortening velocity, SO fascicle length changes during stance, and operating length during running. Significance level was set to ($\alpha \leq 0.05$) and statistical power ($1 - \beta$) was set to 0.8. The estimated sample size required to show significant differences was calculated to 9 participants for fascicle shortening velocity in gastrocnemius muscle. For difference in fascicle length of the GM, the required sample size was calculated to 12 participants.

Results

Muscle-tendon behavior during diagonal stride xc-skiing

Phase characteristics

Average cycle time for each condition can be found in Table 3. The kick phase in DIA_{ref} (which is the phase from zero to the first vertical line in Figure 12 accounted for $15.9 \pm 1.7\%$ of the cycle length. The swing phase, which is the phase from the first vertical line to the second vertical line (Figure 12), accounted for $42.3 \pm 2.5\%$ of the cycle, and the glide phase (second vertical line to end in Figure 12) accounted for the last $44 \pm 1.8\%$ of the cycle.

MTU behavior

There is a stretch-shortening cycle in the MTU during the kick phase with a significant stretch of the MTU in the first part 9.91 ± 0.29 mm (P<.0001), followed by a significant shortening of the MTU 24.29 ± 7.79 mm (P<.0001) (Figure 12). With a net shortening of 14.4 ± 8.7 mm (P<.0001) during the entire kick phase. Furthermore, there is a significant stretch of the MTU in the late glide phase 5.81 ± 0.36 mm (P<.0001). Even though there is a significant shortening throughout the entire glide phase of 10.76 ± 4.27 mm (P<.0001).

There is a stretch and shortening happening in the EE during the kick phase (Figure 12) with a significant stretch in the first part of the kick phase 14.18 ± 5.66 mm (P<.0001), and a significant shortening in the late part of the kick phase 18.56 ± 7.59 mm (P<.0001). With a net shortening during the kick phase of 4.37 ± 7.93 (P=.0351).
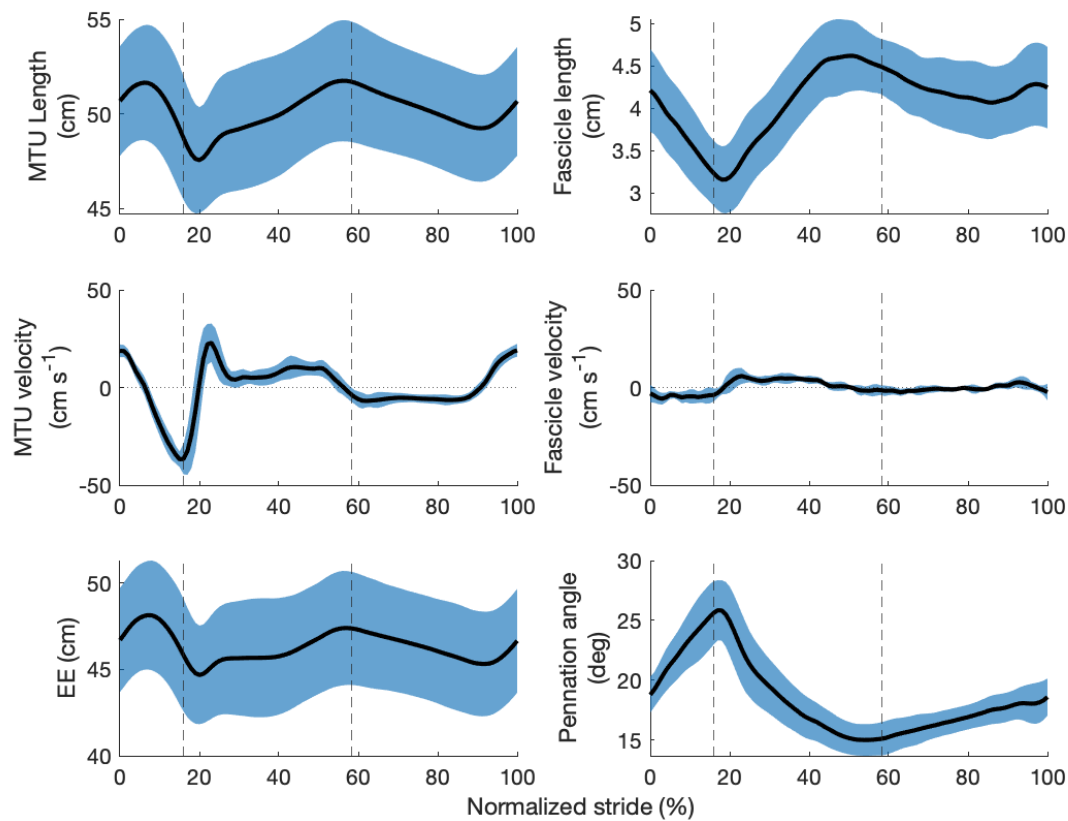
Figure 12 Gastrocnemius medialis (GM) muscle-tendon unit (MTU) length, GM fascicle length, GM MTU velocity, GM fascicle velocity, length of the GM elastic elements (EE), and pennation angle during diagonal stride cycle. Solid lines are the group means. The shaded area is the ± SD. The first stippled vertical line represents the end of the kick phase and the second vertical line represents the start of the glide phase. The dotted vertical line in MTU and fascicle velocity represents zero velocity. Time-series are normalized to 101 points.

Fascicle behavior

The GM muscle fascicle showed a significant shortening during the kick phase of 9.1 ± 2.7 mm (P=<.0001) (Figure 12) with a mean fascicle shortening velocity of 44.8 ± 13.1 mm s^{-1} (P=<.0001), showing a significant concentric muscle contraction in the GM muscle during the kick phase. There was no significant fascicle shortening during the last part of the glide phase. Pennation angle increased during kick $6.4° \pm 1.8°$degrees (P=<.0001) and it increased during the last part of the glide phase $1.8° \pm 1.3°$degrees (P=.0002)

The GM fascicle shortens throughout the entire kick phase, whereas the MTU stretches in the first part of the kick phase and shortens in the later part. This can be seen in Figure 12 where MTU velocity in the first part is positive (meaning MTU lengthens) and crosses zero and becoming negative in the later part, meaning the MTU shortens. The muscle fascicles are constantly shortening throughout the kick phase, but the shortening velocity of the fascicle is

relatively low compared to the shortening velocity of the MTU, with a peak fascicle shortening of 73.8 ± 32.0 mm s^{-1} compared to the peak shortening velocity of the MTU of 375.3 ± 64.8 mm s^{-1}.

Muscle activation

During the kick phase, both muscle activation of the GL and SO muscle was significantly active based on a threshold value of 25%, with a mean EMG activity level at 66.7 ± 11.7 % of EMG$_{max}$ and 65.8 ± 4.0 % of EMG$_{max}$ for the GL and SO muscle respectively. During the late part of the glide phase, we found no significant muscle activation in the GL muscle. However, we found a significant muscle activation in the SO muscle in this phase 29.3 ± 7.0 % of EMG$_{max}$ (P=.0227) (Figure 13).
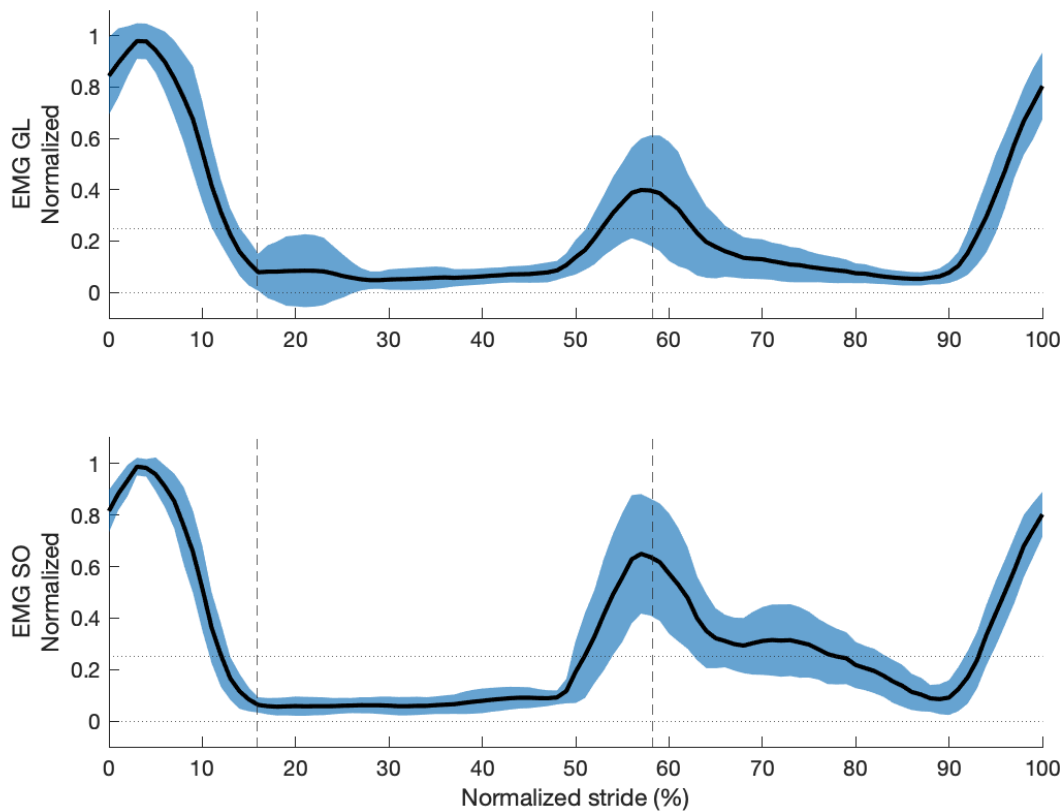

Figure 13 Electromyographic (EMG) activity of gastrocnemius lateralis (GL) and soleus (SO) during diagonal stride cycle. Solid lines are the group means, and the shaded area is ± SD. EMG is normalized to peak muscle activity during DIA$_{ref}$. Time series are normalized to 101 points.

Qualitatively we can see that the force is above 1 which equals to body mass for a major part of the kick phase (Figure 14). As expected, the force is equal to zero in the swing phase when there is no contact between the ski and the ground. In the glide phase, the force is around 1 or equal to body mass. During the late part of the glide phase there appear to be a decline of force with an average minimum value 0.8 ± 0.4, prior to an increase in force with an average maximum value of 1.4 ± 0.6.
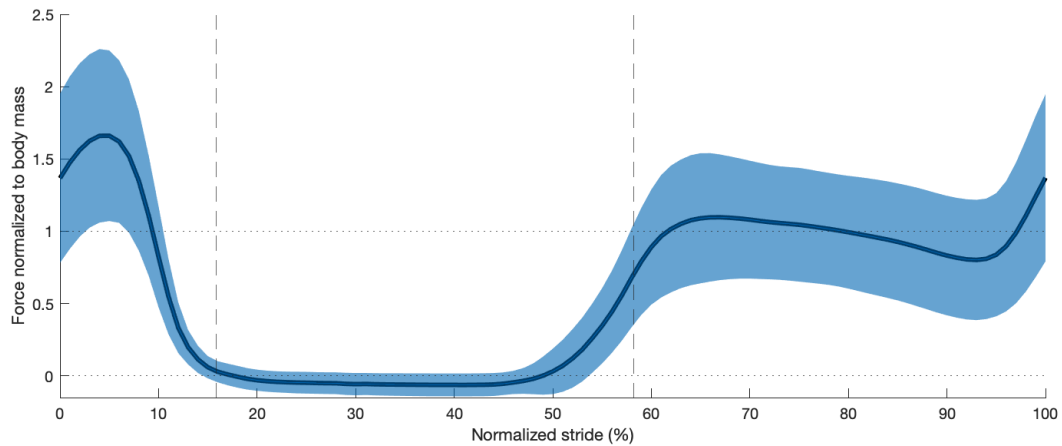

Figure 14 Normalized Force in ski. The solid line is the group mean. The shaded area is ± SD. The first stippled vertical lines represent the end of the kick phase and the second represent the start of the glide phase. The dotted horizontal line starting in 1 represents the body mass, the dotted line starting at zero represents zero force. Force measurements are normalized to body weight. Time series are normalized to 101 points

Effect of increased speed and incline on muscle-tendon behavior during diagonal stride xc-skiing

MTU behavior

There was a significant effect of both speed and incline on the stretch of the MTU during the kick phase, with the peak length being higher in both the DIA_{fast} and DIA_{steep} compared to the DIA_{ref} (Figure 15). The peak length of MTU during the kick phase in DIA_{ref} was 516.7 ± 30.4 mm. In DIA_{fast}, the peak MTU length increased to 518.6 ± 31.4 mm, with a mean difference of 1.8 mm (P=.0022). Peak MTU length increased to 522.2 ± 31.8 mm in DIA_{steep}, and the mean difference compared to DIA_{ref} was 5.4 mm (P<.0001).

EE followed a similar trend as the MTU. EE peak length during the kick phase in DIA_{ref} was 481.7 ± 31.4 mm. In DIA_{fast} the peak length increased by 2.17 mm to 483.9 ± 32.6 mm ($P=.0172$). In DIA_{steep} peak EE length increased to 488.7 ± 33.5 mm, an increase of 7.1 mm ($P<.0001$).
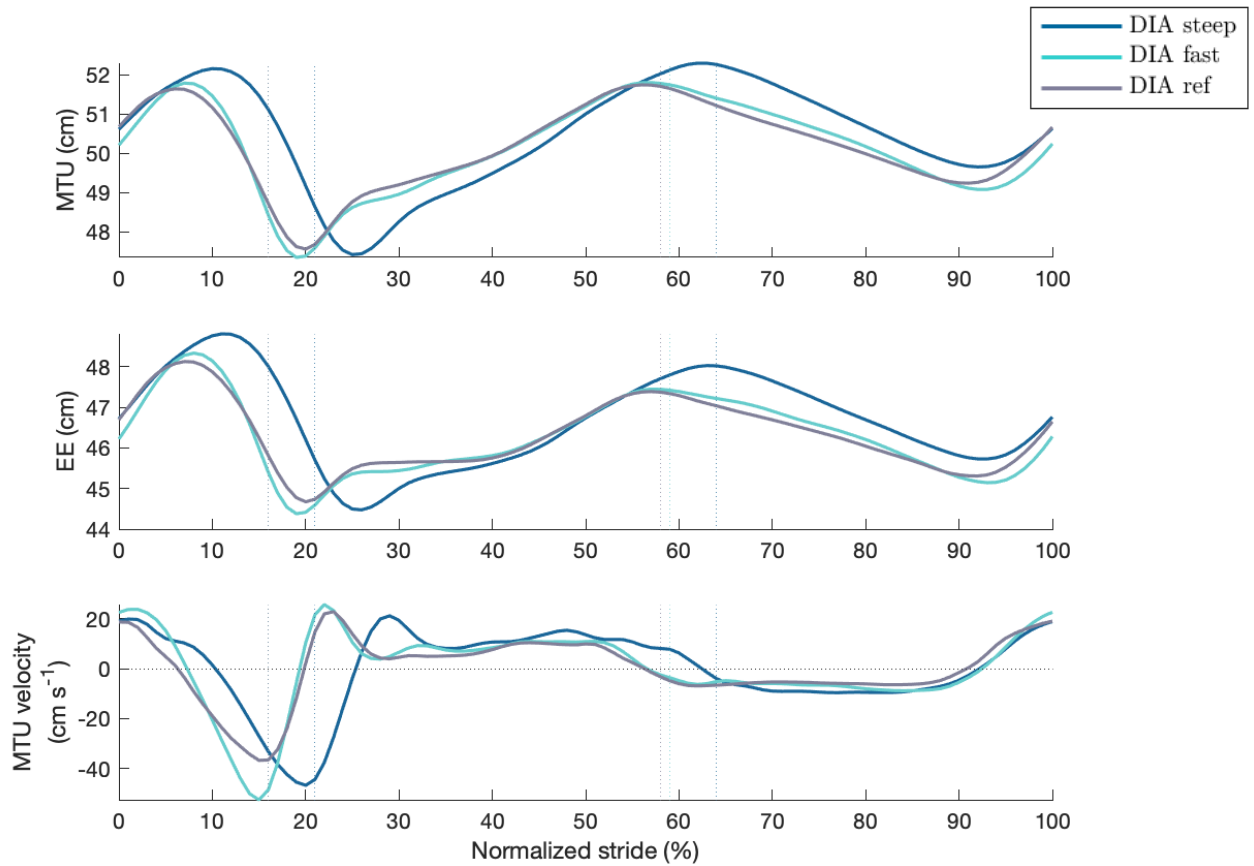
Figure 15 Gastrocnemius medialis (GM) muscle-tendon unit (MTU) length, length of elastic elements (EE), and MTU velocity during diagonal stride under three different conditions (DIA_{ref}, DIA_{fast}, and DIA_{steep}). Dotted vertical lines show the end of the kick phase and the start of the glide phase for the different conditions. Vertical lines indicate the end of the kick phase and the start of the glide phase. Time-series is normalized to 101 points

The MTU length changes during the part of the kick phase where MTU stretches. Increased significantly with both increased speed and increased incline. The length change of MTU during DIA_{ref} was 9.91 ± 2.88 mm. In DIA_{fast}, MTU length change was 16.4 ± 6.36 mm, and the mean difference between DIA_{ref} and DIA_{fast} increased by 6.49 mm ($P=.0003$). The length change of the MTU during DIA_{steep} was 16.0 ± 6.23 mm, thus having a mean difference 6.1 mm compared to DIA_{ref}.

EE follow a similar trend as MTU. The length changes of EE in DIA_{ref} during MTU stretch were, 14.2 ± 2.9 mm. Length changes of the EE in DIA_{fast} were 21.3 ± 7.8 mm with a mean difference from DIA_{ref} of 7.1 mm (P=.0019). The length change of EE during DIA_{steep} was 21.2 ± 7.9 mm, with a mean difference from the DIA_{ref} of 7.0 mm (P=.0031).

Even though MTU showed significantly different length changes during the MTU stretch phase as a factor of speed and incline, MTU did not show differences in length changes during MTU shortening.

Length change in EE during MTU shortening for DIA_{ref} was 18.6 ± 7.6mm. In DIA_{fast}, the EE length change during MTU shortening was 21.58 ± 11.1mm, which was not significantly different from the length changes in DIA_{ref}. EE length change in DIA_{steep} was 26.1 ± 14.8mm, thus the mean difference between DIA_{ref} and DIA_{steep} was 7.6 mm (P=.0473).

Fascicle behavior

Mean fascicle shortening velocity of GM during kick does not significantly differ between DIA_{ref} and DIA_{fast}, with a mean shortening velocity of 44.8 ± 13.1 mm s^{-1} and 45.6 ± 17.0 mm s^{-1} for the DIA_{ref} and DIA_{fast} respectively (Figure 16). The mean fascicle shortening velocity of GM during kick during DIA_{steep} was significantly lower from the fascicle shortening during DIA_{ref} with a shortening velocity of 34.2 ± 14.8 mm s^{-1}, and the mean difference 10.6 mm s^{-1} (P=.0007).

There were no significant differences in peak GM fascicle shortening velocity during kick between DIA_{ref} and DIA_{fast}, with peak fascicle shortening velocities of 73.8 ± 32.0 mm s^{-1} and 78.3 ± 38.1 mm s^{-1} for the DIA_{ref} and DIA_{fast}. The peak shortening velocity of GM in DIA_{steep} was significantly lower from the peak fascicle shortening velocity of DIA_{ref}. With peak fascicle shortening velocity in DIA_{steep} of 65.9 ± 29.8 mm s^{-1} and the mean difference of DIA_{ref} and DIA_{steep} of 7.9 mm s^{-1} (P=.0346).

Mean fascicle length did not vary significantly across the conditions. Length changes in the fascicle length were significantly different between DIA_{ref} and DIA_{steep}, with mean length change for DIA_{ref} of 9.1 ± 2.7 mm and the mean length change DIA_{steep} of 7.6 ± 3.2 mm. Given a mean difference of 1.5 mm (P=.027). Fascicle length change between DIA_{ref} and DIA_{fast} was not significant.
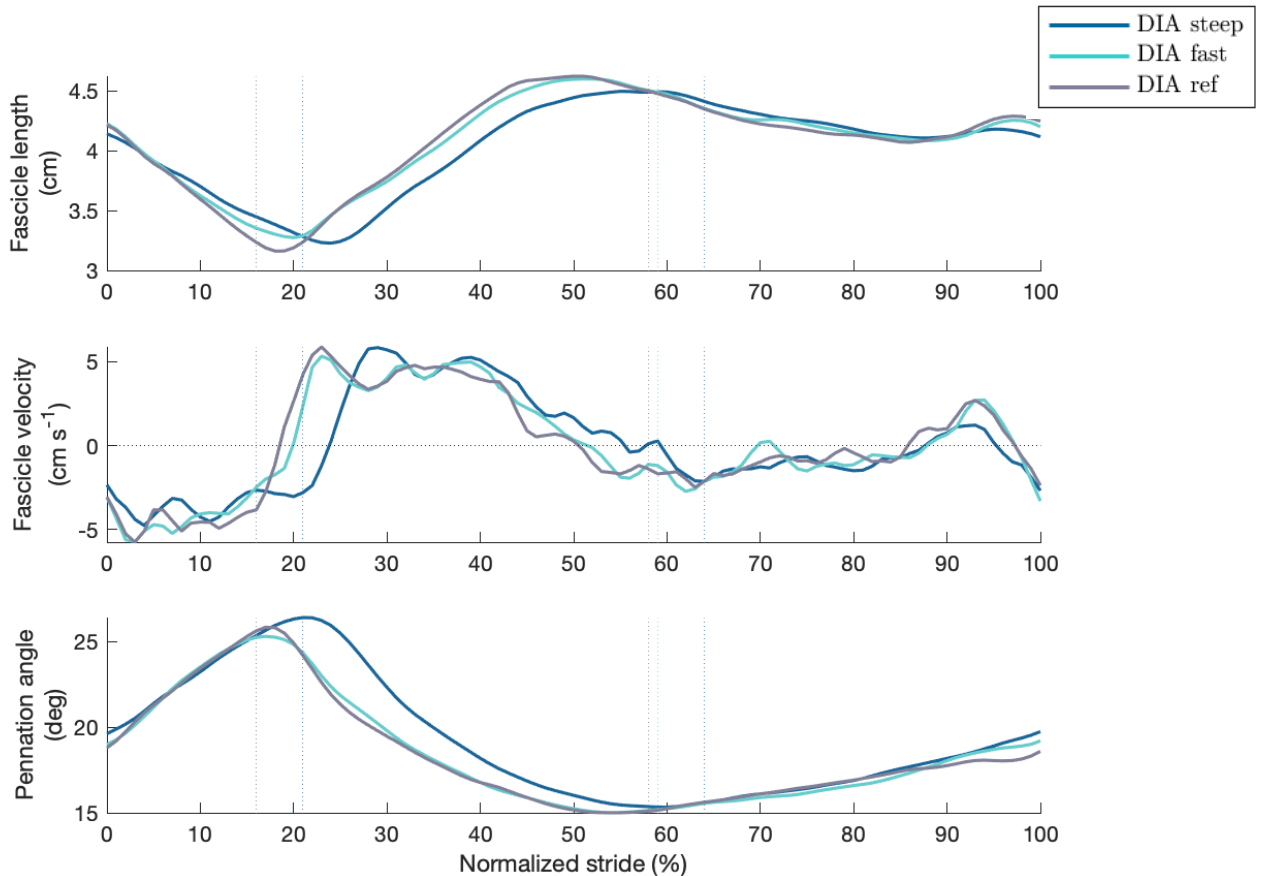
Figure 16 Gastrocnemius medialis (GM) fascicle length, fascicle velocity, and pennation angle during diagonal stride under three different conditions; DIA_{ref}, DIA_{fast}, DIA_{steep}. Dotted vertical lines show the end of the kick phase and the start of the glide phase for the different conditions. Time-series is normalized to 101 points.

Muscle activity

Increase in speed had no significant effect on the integrated EMG (iEMG) for the GL and the SO muscle during the kick phase. However, there was an effect of incline on iEMG, with DIA_{steep} showing increased muscle activity compared to DIA_{ref}. iEMG for GL was 0.14 ± 0.02 for DIA_{ref} vs. 0.22 ± 0.03 for DIA_{steep} with a mean difference 0.07 (P<.0001). For SO the iEMG increased from 0.14 ± 0.01 in the DIA_{ref} to 0.20 ± 0.02 in DIA_{steep} (P<.0001).

Both speed and incline affected peak muscle activity during the kick phase (Figure 17). The peak muscle activity increased with 23.1% (P=.0006) for the GL muscle and 10.7%(P=0.025) for the SO muscle with increased speed. With increased incline, the peak muscle activity increased with 23.8% (P=0.037) in the GL muscle, and with 15.4% (P=.0001) in the SO muscle.
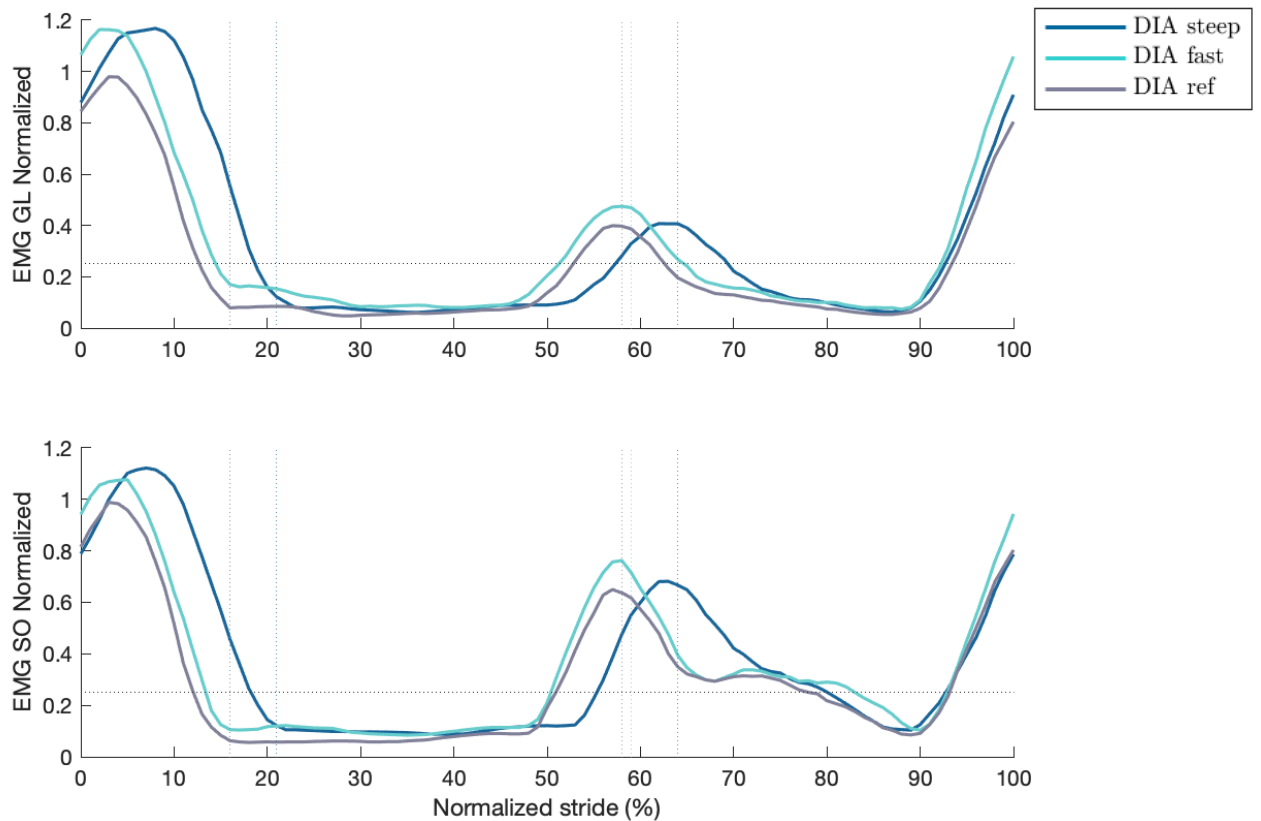
Figure 17 Electromyographic (EMG) activity of gastrocnemius lateralis (GL) and soleus (SO) during the diagonal stride. Data er mean for three different conditions (DIA$_{ref}$, DIA$_{fast}$, and DIA$_{steep}$). Vertical dotted lines indicate the end of the kick phase and the start of the glide phase. Time series are normalized to 101 points.

Cycle characteristics

The cycle time for each condition as well as the duration of the kick, swing, and glide phase can be found in Table 3. Cycle time gets significantly shorter with both increased speed and increased incline, with a mean difference 0.12 ± s (P<.0001) and 0.18 s (P<.0001) for increased speed and increased incline respectively.

Table 3 Cycle and phase time for the diagonal stride during each condition

	Cycle time (s)	Time kick (s) (% of cycle)	Time swing (s) (% of cycle)	Time glide (s) (% of cycle)
DIA$_{ref}$	1.35 ± 0.05	0.21 ± 0.03 (15.9%)	0.57 ± 0.04 (42.4%)	0.59 ± 0.03 (44%)
DIA$_{fast}$	1.23 ± 0.07*	0.19 ± 0.04* (15.6%)	0.53 ± 0.04** (43.0%)	0.53 ± 0.03** (43.5%)
DIA$_{steep}$	1.17 ± 0.05*	0.24 ± 0.04* (20.8%)**	0.51 ± 0.04** (43.5%)	0.44 ± 0.03** (37.8%)**

Values are mean ± SD; s = seconds; DIA$_{ref}$ = 2.5m s^{-1}, 5°; DIA$_{fast}$ = 3.5 m s^{-1}, 5°; DIA$_{steep}$ = 2.5 m s^{-1}, 10°; * = significantly different from DIA$_{ref}$ (P<.05); ** = significantly different from DIA$_{ref}$ (P<.01).

The duration of the kick phase gets significantly shorter in DIA$_{fast}$ with a mean difference of 0.02s (P=.0332). whereas the duration of the kick phase increases with increased incline, mean difference 0.03s (P=.0101).

Percentage wise there is no significant difference in the relative duration of the kick (compared to the entire cycle) between the DIA$_{ref}$ and DIA$_{fast}$. However, at an increased incline there is a significant increase in the percentage wise duration of the kick phase with a mean increase of 4.9% (P<.0001)

Duration of the glide phase is significantly shorter for both DIA$_{fast}$ and DIA$_{steep}$ compared to DIA$_{ref}$ with a mean difference of 0.06 s (P<.0001) and 0.15 s (P<.0001) for DIA$_{fast}$ and DIA$_{steep}$, respectively. There is no significant difference percentage wise between the DIA$_{ref}$ and DIA$_{fast}$ but there is a significant decrease in the percentage wise duration of the glide phase in DIA$_{steep}$ compared to DIA$_{ref}$ with a mean difference of 6.2% (P<.0001).

Swing time is significantly lower for both DIA$_{fast}$ and DIA$_{steep}$, with a mean difference of 0.04 (P=.0029) and 0.06 (P=.0002) respectively. However, there are no significant changes percentage wise meaning that the change in cycle time came from differences in the kick and glide phase, whereas the percentage swing time remains similar during the different conditions.

Discussion

In this study, we investigated the gastrocnemius muscle-tendon mechanics during DIA on roller skies. We combined ultrasound measurements with kinematic and kinetic measurements to better understand the mechanics of the GM MTU and how it is affected by external condition i.e., speed and incline. The main findings in this study are that fascicles shorten throughout the entire kick phase, whereas the MTU undergoes a stretch-shortening cycle. Thus, elastic energy can be stored in the EE and then later be released during the kick phase. Further, MTU lengthens in the last part of glide, whereas the fascicles remain isometric. However, data on whether the muscle was active in the last part of glide were inconclusive, since SO muscle was active, whereas GL muscle was not. MTU stretch increased with both increased speed and incline, likely because the MTU modulates it function when under different time constraint. Lai et al., (2014) and Monte et al., (2020) found that as running speed increases so does the relative contribution of tendon elastic strain to the positive work done by the MTU, thus, the tendon stretch is increased. Werkhausen et al., (2019) found that at increased load (which is comparable to increased incline in this study) allowed fascicles to operate at lower shortening velocities and therefore contact under favorable conditions to produces increased force, which may explain the increased stretch of the MTU and EE.

Muscle-tendon behavior and biomechanics during diagonal stride XC skiing

The main hypothesis was that the propulsion was generated by concentric muscle contraction in the kick phase, while the MTU and EE undergoes a stretch-shortening cycle during the kick phase. The data confirmed this with the fascicle continually shortening throughout the entire duration of the kick. Further, the data show that the MTU undergoes a stretch followed by a shortening during the kick phase, and the same is true for the EE (Figure 12).

The data shows that the MTU length changes and fascicle length changes are decoupled from each other with MTU lengthening in the first part of the kick phase before shortening (recoil) in the later part of the kick phase, whereas the fascicle shortens throughout the entire kick phase. This can also be seen in the velocity of the fascicles compared to the velocity of the MTU. The fascicles have a negative shortening velocity through the entire kick phase (i.e., continuous shortening of the fascicles). Whereas, the MTU has a positive shortening velocity (i.e., lengthening) in the first part, before having a negative shortening velocity in the later

part (i.e., shortening). This is in line with other studies done on running (Lichtwark et al., 2007). The shortening of the fascicles throughout the kick phase implies that energy generated by the muscles are stored within the EE, as MTU lengthens in the first part of the kick phase. Before this energy is utilized in the late part of the kick phase as the MTU and EE shorten rapidly.

The peak MTU shortening velocity is five times higher than the peak velocity of the muscle fascicles. (375.3 vs. 73.8 mm s^{-1}). These findings are in line with Lichtwark and Wilson, (2006), who found that the rate of shortening in the fascicle was typically 25-30% of the total MTU shortening velocity in running. This means that the elastic recoil of the tendon and aponeurosis contributed most to MTU shortening. Therefore, fascicles could operate at a relatively low shortening velocity, allowing the fascicles to work at a more favorable part of the F-V curve to produce greater force economically. During the kick, the shortening length changes are three times larger in the MTU compared with the fascicle (24.3 vs 9.1 mm for the MTU and fascicles respectively).

Furthermore, it was hypothesized that muscle contraction was initiated during the last part of the glide phase, and that there is a preload phase prior to the kick, as proposed by Komi and Norman (1987), where elastic energy could be stored. This has been described as an unweighting phase prior to the kick, and should be visual in the force curve as a decrease in force shortly before the kick. There was a significant lengthening of the MTU and the EE in the last part of glide, with muscle fascicles remaining relative isometric, indicating that there may be a buildup of pretension (and possibly storage of elastic energy) in this phase. To store elastic energy, the muscle should be active during the lengthening of the MTU and EE. The SO muscle was significantly active during the last part of the glide phase, whereas the GL was not. Furthermore, there appears to be a drop in the force curve during the late glide phase (Figure 3), however, this was not always present at the individual level. This could be due to either individual differences or due to limitations in our force measurements. Therefore, no conclusive evidence of muscle activation and storage of elastic strain energy in the last part of glide was found.

The EMG seems to traverse the threshold for muscle activation (Özgünen et al., 2010) only in the last ≈ 6-7% (Figure 13) of the cycle and thus not be picked up due to the method of phase

division. However, pennation angle increased in the last part of the glide phase. This may indicate that the muscle is active since increase in pennation angle requires muscle activation.

Effect of speed and incline on muscle-tendon behavior during XC skiing diagonal stride

Our second hypothesis is that increased speed and incline influence the muscle-tendon behavior. We believed that higher speed would increase EMG activation, fascicle shortening velocity, and stretch of the MTU and EE, as well as affect cycle characteristics by decreasing the kick time. With increased incline we expected increased muscle activation, similar fascicle shortening velocity, increased stretch of the EE and EMG, as well as different cycle characteristics by decreasing overall cycle time. How speed and incline affected each variable is described in the sections below.

Cycle time and phase characteristics

Cycle time is significantly affected by both speed and incline (Table 3). If we look at the phase characteristics as a percentage of the cycle, DIA_{ref} and DIA_{fast} are similar, meaning DIA_{fast} is comparable to DIA_{ref}, just with a shorter overall cycle time, which aligns with our hypothesis about the different time constraints. DIA_{steep} has a shorter cycle time and different phase characteristics than DIA_{ref}. Even though cycle time is shorter in DIA_{steep}, the kick phase is longer both in absolute time (s) and relative time (% of cycle), compared with DIA_{ref}. The increased length of the kick phase is likely due to the greater work demand when increasing the body´s potential energy at a higher incline, this is also evident in a significant increase of iEMG. The longer time period to generate force may benefit the contractile conditions, allowing fascicles to produce force over a longer period.

MTU behavior

Both increased speed and incline had a significant effect on the length of MTU and EE during the kick phase, with a peak length of MTU and EE being higher in both. The MTU and EE lengths were the highest in DIA_{steep}, although the length change in DIA_{fast} was higher. The longer MTU and EE in DIA_{steep} length is likely because of the more dorsi flexed ankle caused by the increased incline. MTU and EE length changes were highest in DIA_{fast}, indicating a possible increase in utilization of stored elastic energy with advancing speed. Several studies have demonstrated that utilization of elastic strain energy by the tendons can

optimize the region where muscle fascicle operate on the F-L and F-V curve at advancing running speeds, and that the elastic strain energy stored when the Achilles tendon stretches and recoils increases as function of running speed (Dorn et al., 2012; A. Lai et al., 2014; Monte et al., 2020). Lai et al., (2014) reported the contribution of the tendon elastic strain energy to the positive work generated by the MTU to increase from 62% to 75% in the gastrocnemius when running speed increased. Monte et al., (2020) also found that the mechanical power provided by the EE during the propulsive phase in running increased as a function of speed. Thus, the increase in MTU and EE length changes in DIA_{fast} found in this study may indicate increased utilization of elastic strain energy, since there were no significant differences in fascicle behavior to explain the increase in force. Thus, the MTU may prioritize the storage and recoil of elastic strain energy over muscle fascicle work as speeds advances also in DIA XC-skiing.

Fascicle behavior

Contrary to our hypothesis mean fascicle shortening velocity was not significantly affected by increased speed. However, the mean fascicle shortening velocity in DIA_{steep} was significantly lower than the fascicle shortening velocity of DIA_{ref}. This is in contrast to the hypothesis that the fascicle shortening velocities should be similar. However, this is in line with a study by Werkhausen et al., (2019) who studied the triceps surae muscle-tendon interaction with increased requirements for force and work during running by increasing both speed and load (with a weighted west). They found that when ground contact could be prolonged (i.e., with increased load) fascicle contraction velocity was preserved or lowered because force was produced over a longer time period. These findings indicate that the neuromuscular system meets the increased mechanical demand by favoring economical force production when enough time is available. Thus, the increased length of kick phase in DIA_{steep} allows muscle fascicle to contract at lower shortening velocities, favoring economical force production to meet the increased force requirements of steeper incline.

Fascicle length change in DIA_{steep} is also significantly lower than in DIA_{ref}, whereas there is no difference in length change between DIA_{ref} and DIA_{fast}. This may be because as incline increases so does the contribution of other joints such as the knee and hip joint (Roberts & Belliveau, 2005). Thus, more of the work is produced by muscles in proximal joints, decreasing length changes in the GM muscle. Furthermore, Lichtwark and Wilson (2006)

suggested that the muscle fascicles suggested that the muscle fascicles contract at higher velocity with increased incline, which would requiring more activation to produce the same force according to the F-V relationship, or that this could be caused by variation in force sharing between the muscles of the triceps surae.

Muscle activation

Contrary to our hypothesis, the iEMG for GL and SO did not significantly increase from DIA_{ref} to DIA_{fast}. This could be because the relative contribution of the tendon increased when speed was higher, which is the case in running (Cavagna, 2009; A. Lai et al., 2014; Monte et al., 2020), or it could be that the decrease in kick time affected the iEMG, which is also time dependent. Between DIA_{ref} and DIA_{steep} there was a significant increase in iEMG, which was in line with our hypothesis. The increase of iEMG in DIA_{steep} is likely due to the increased demand for net mechanical work with skiing on an increased incline, and indicates that an greater CSA of the muscle is recruited to meet this demand (Roberts & Belliveau, 2005). Furthermore, since iEMG is the time integral of the EMG curve, the slightly longer kick phase may also contribute to the increased iEMG in DIA_{steep}.

Even though iEMG was similar between DIA_{ref} and DIA_{fast}, Peak EMG increased significantly in both GL and SO. With an increase of 23.1% in GL and 10.7% in SO. Peak EMG increased in DIA_{steep} also, with a 23.8% increase in GL and 15.4% in SO.

Limitations

Although the advancements have been significant in recent decades, there are still methodical limitations when studying muscle-tendon dynamics in vivo using US imaging. Perhaps the most significant methodical issue regarding US imaging of the muscles, is that it relies primarily on a conceptual model that assumes the muscle fascicles to act in a 2D plane. This assumption that muscle fascicles act in the same 2D plane as the ultrasound imaging during the stance phase, gloss over the complex 3D changes of the muscle during contractions (Lai et al., 2015; Roberts et al., 2019).

The US probes were secured to the body using a custom-made bracket and elastic bands. Although this set up was quite efficient at keeping the probe in place, some movements of the probe are expected. The elastic bands may compress the muscle. Thus, it might affect the 3D

changes in the muscle, i.e., how the muscle bulges. Thus, affecting the lateral forces in the muscle.

In this study we used a conceptional model from Fukunaga et al. (2001) that assumes tendon and EE work in series along the length of the muscle. However, fibers within the tendons can twist and curve and may therefore not strain from point to point. We might also have contributions from elastic tissue that act in a different direction than the muscle fascicles when stretched, such as aponeurosis and the perimysium of the muscles (Lai et al., 2015; Lichtwark et al., 2007).

Reflective markers on the foot (RHEEL, RTOE 1, 2, 5) and in most cases the ankle (RLM, RMM) (some skiers had low cut ski boots with visible malleolus) was placed on the outside of the shoe. Therefore, the reflective marker is not fixed to the anatomical landmark and the foot may move inside the shoe, so that the anatomical landmarks doesn´t align with its respective markers, since these markers is used to estimate the ankle angle, this can affect the calculation of MTU length.

Force measurements were done using two FSRs that had a specified load range 0-445N, which is lower than the expected forces, therefore, we believed the force data to suffer from lack of linearity when force exceeded the specified load range. Even though our force data seemed sensible, it was only used to locate the start of the glide phase. The force measurements were useful in helping to understand how the different phases related to the skiing technique.

Skiing on roller skies and skiing on snow have some differences which should be considered. During diagonal stride on snow the skier needs to push down the ski camber so that the wax can grip the snow. On the contrary, with the ratchet wheel mechanism in roller skies you have almost unlimited grip on the surface, thus different muscle-tendon behavior is possible. Despite the challenges we believe that roller skiing is a good substitute for skiing on snow. In a review by Zoppirolli et al., (2020), they found that at least half of the studies on XC-skiing were conducted on a treadmill.

Conclusion

This study showed that during the kick phase of DIA XC-skiing the GM muscle fascicle shortens continuously whereas the MTU lengthens in the first part of the kick phase allowing for storage of elastic energy, which is released later in the kick phase when the MTU shortens rapidly. Thus, fascicle behavior is decoupled from that of the whole MTU. This may allow the fascicle to operate at lower shortening velocities and at a more favorable fascicle lengths to produce high force economically. Based on the data we could not conclude whether there is storage of elastic energy being initiated in the glide phase.

Peak MTU length increased with both increased speed and incline. The length changes of the MTU were highest in DIA_{fast} indicating that more elastic energy was stored with increased speed. Fascicle shortening velocity did not increase with higher speeds, however, at increased incline the fascicle shortening velocity decreased, indicating that fascicles work at a more favorable part of the F-L-V relationship. iEMG increased in DIA_{steep} indicating that more muscle activations were necessary to accommodate the increased power demand.

References

Aagaard, P., Andersen, J. L., Dyhre-Poulsen, P., Leffers, A.-M., Wagner, A., Magnusson, S. P., Halkjær-Kristensen, J., & Simonsen, E. B. (2001). A mechanism for increased contractile strength of human pennate muscle in response to strength training: Changes in muscle architecture. *The Journal of Physiology*, *534*(Pt 2), 613–623. https://doi.org/10.1111/j.1469-7793.2001.t01-1-00613.x

Alexander, R. M., & Vernon, A. (1975). The mechanics of hopping by kangaroos (Macropodidae). *Journal of Zoology*, *177*(2), 265–303. https://doi.org/10.1111/j.1469-7998.1975.tb05983.x

Arnold, E. M., Hamner, S. R., Seth, A., Millard, M., & Delp, S. L. (2013). How muscle fiber lengths and velocities affect muscle force generation as humans walk and run at different speeds. *Journal of Experimental Biology*, *216*(11), 2150–2160. https://doi.org/10.1242/jeb.075697

Biewener, A. A., & Roberts, T. J. (2000). Muscle and tendon contributions to force, work, and elastic energy savings: A comparative perspective. *Exercise and Sport Sciences Reviews*, *28*(3), 99–107.

Bohm, S., Marzilger, R., Mersmann, F., Santuz, A., & Arampatzis, A. (2018). Operating length and velocity of human vastus lateralis muscle during walking and running. *Scientific Reports*, *8*(1), 1–10. https://doi.org/10.1038/s41598-018-23376-5

Bohm, S., Mersmann, F., Santuz, A., & Arampatzis, A. (2019). The force–length–velocity potential of the human soleus muscle is related to the energetic cost of running. *Proceedings of the Royal Society B: Biological Sciences*, *286*(1917), 20192560. https://doi.org/10.1098/rspb.2019.2560

Cavagna, G. A. (2009). The two asymmetries of the bouncing step. *European Journal of Applied Physiology*, *107*(6), 739. https://doi.org/10.1007/s00421-009-1179-2

Challis, J. H. (2000). Muscle-Tendon Architecture and Athletic Performace. In V. Zatsiorsky, *Biomechanics in Sport: Performance Enhancement and Injury Prevention*. Oxford: Wiley-Blackwell. http://web.a.ebscohost.com/ehost/ebookviewer/ebook/bmxlYmtfXzIzMzIyNF9fQU41 ?sid=b0bdef21-c51d-4bcd-9a06-c4bdf752873c@sdc-v-sessmgr01&vid=0&format=EB&rid=1

Cronin, N. J., Avela, J., Finni, T., & Peltonen, J. (2013). Differences in contractile behaviour between the soleus and medial gastrocnemius muscles during human walking. *Journal of Experimental Biology*, *216*(5), 909–914. https://doi.org/10.1242/jeb.078196

Cronin, N. J., Carty, C. P., Barrett, R. S., & Lichtwark, G. (2011). Automatic tracking of medial gastrocnemius fascicle length during human locomotion. *Journal of Applied Physiology*, *111*(5), 1491–1496. https://doi.org/10.1152/japplphysiol.00530.2011

Dorn, T. W., Schache, A. G., & Pandy, M. G. (2012). Muscular strategy shift in human running: Dependence of running speed on hip and ankle muscle performance. *Journal of Experimental Biology*, *215*(11), 1944–1956. https://doi.org/10.1242/jeb.064527

Enoka, R. M. (2015). *Neuromechanics of human movement* (fifth). Human Kinetics.

Ettema, G. (2001). Muscle efficiency: The controversial role of elasticity and mechanical energy conversion in stretch-shortening cycles. *European Journal of Applied Physiology*, *85*(5), 457–465. https://doi.org/10.1007/s004210100464

Farris, D. J., & Lichtwark, G. A. (2016). UltraTrack: Software for semi-automated tracking of muscle fascicles in sequences of B-mode ultrasound images. *Computer Methods and Programs in Biomedicine*, *128*, 111–118. https://doi.org/10.1016/j.cmpb.2016.02.016

Farris, D. J., & Sawicki, G. S. (2012a). The mechanics and energetics of human walking and running: A joint level perspective. *Journal of The Royal Society Interface*, *9*(66), 110–118. https://doi.org/10.1098/rsif.2011.0182

Farris, D. J., & Sawicki, G. S. (2012b). Human medial gastrocnemius force–velocity behavior shifts with locomotion speed and gait. *Proceedings of the National Academy of Sciences*, *109*(3), 977–982. https://doi.org/10.1073/pnas.1107972109

Faul, F., Erdfelder, E., Buchner, A., & Lang, A.-G. (2009). Statistical power analyses using G*Power 3.1: Tests for correlation and regression analyses. *Behavior Research Methods*, *41*(4), 1149–1160. https://doi.org/10.3758/BRM.41.4.1149

Faul, F., Erdfelder, E., Lang, A.-G., & Buchner, A. (2007). G*Power 3: A flexible statistical power analysis program for the social, behavioral, and biomedical sciences. *Behavior Research Methods*, *39*(2), 175–191. https://doi.org/10.3758/BF03193146

Fenn, W. O. (1924). The relation between the work performed and the energy liberated in muscular contraction. *The Journal of Physiology*, *58*(6), 373–395. https://doi.org/10.1113/jphysiol.1924.sp002141

FIS. (2020). *International Competition Rules Cross-Country*. Cross-Country Documents. https://www.fis-ski.com/en/inside-fis/document-library/cross-country-documents

Frontera, W. R., & Ochala, J. (2015). Skeletal Muscle: A Brief Review of Structure and Function. *Calcified Tissue International*, *96*(3), 183–195. https://doi.org/10.1007/s00223-014-9915-y

Fukunaga, T., Kubo, K., Kawakami, Y., Fukashiro, S., Kanehisa, H., & Maganaris, C. N. (2001). In vivo behaviour of human muscle tendon during walking. *Proceedings of the Royal Society of London. Series B: Biological Sciences*, *268*(1464), 229–233. https://doi.org/10.1098/rspb.2000.1361

Gans, C., & Gaunt, A. S. (1991). Muscle architecture in relation to function. *Journal of Biomechanics*, *24*, 53–65. https://doi.org/10.1016/0021-9290(91)90377-Y

Gordon, A. M., Huxley, A. F., & Julian, F. J. (1966). The variation in isometric tension with sarcomere length in vertebrate muscle fibres. *The Journal of Physiology*, *184*(1), 170–192. https://doi.org/10.1113/jphysiol.1966.sp007909

Hawkins, D., & Hull, M. L. (1990). A method for determining lower extremity muscle-tendon lengths during flexion/extension movements. *Journal of Biomechanics*, *23*(5), 487–494. https://doi.org/10.1016/0021-9290(90)90304-L

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology*, *10*(5), 361–374. https://doi.org/10.1016/S1050-6411(00)00027-4

Herzog, W. (2000). Mechanical properties and performance in skeletal muscles. In V. Zatsiorsky, *Biomechanics in Sport: Performance Enhancement and Injury Prevention*. Oxford: Wiley-Blackwell. http://web.a.ebscohost.com/ehost/ebookviewer/ebook/bmxlYmtfXzIzMzIyNF9fQU41?sid=b0bdef21-c51d-4bcd-9a06-c4bdf752873c@sdc-v-sessmgr01&vid=0&format=EB&rid=1

Hill, A. V. (1938). The heat of shortening and the dynamic constants of muscle. *Proceedings of the Royal Society of London. Series B - Biological Sciences*, *126*(843), 136–195. https://doi.org/10.1098/rspb.1938.0050

Hill, A. V. (1964). The effect of load on the heat of shortening of muscle. *Proceedings of the Royal Society of London. Series B. Biological Sciences*. https://doi.org/10.1098/rspb.1964.0004

Kehler, A. L., Hajkova, E., Holmberg, H.-C., & Kram, R. (2014). Forces and mechanical energy fluctuations during diagonal stride roller skiing; running on wheels? *Journal of Experimental Biology*, *217*(21), 3779–3785. https://doi.org/10.1242/jeb.107714

Komi, P. V. (1990). Relevance of in vivo force measurements to human biomechanics. *Journal of Biomechanics*, *23*, 23–34. https://doi.org/10.1016/0021-9290(90)90038-5

Komi, P. V., & Norman, R. W. (1987). Preloading of the Thrust Phase in Cross-Country Skiing. *International Journal of Sports Medicine*, *08*(S 1), S48–S54. https://doi.org/10.1055/s-2008-1025703

Lai, A. K. M., Lichtwark, G. A., Schache, A. G., & Pandy, M. G. (2018). Differences in in vivo muscle fascicle and tendinous tissue behavior between the ankle plantarflexors during running. *Scandinavian Journal of Medicine & Science in Sports*, *28*(7), 1828–1836. https://doi.org/10.1111/sms.13089

Lai, A., Lichtwark, G. A., Schache, A. G., Lin, Y.-C., Brown, N. A. T., & Pandy, M. G. (2015). In vivo behavior of the human soleus muscle with increasing walking and running speeds. *Journal of Applied Physiology*, *118*(10), 1266–1275. https://doi.org/10.1152/japplphysiol.00128.2015

Lai, A., Schache, A. G., Lin, Y.-C., & Pandy, M. G. (2014). Tendon elastic strain energy in the human ankle plantar-flexors and its role with increased running speed. *Journal of Experimental Biology*, *217*(17), 3159–3168. https://doi.org/10.1242/jeb.100826

Levangie, P. K., & Norkin, C. C. (2011). *Joint structure and function: A comprehensive analysis* (5th ed.). F. A. Davis Company.

Lichtwark, G. A., Bougoulias, K., & Wilson, A. M. (2007). Muscle fascicle and series elastic element length changes along the length of the human gastrocnemius during walking and running. *Journal of Biomechanics*, *40*(1), 157–164. https://doi.org/10.1016/j.jbiomech.2005.10.035

Lichtwark, G. A., & Wilson, A. M. (2005). In vivo mechanical properties of the human Achilles tendon during one-legged hopping. *Journal of Experimental Biology*, *208*(24), 4715–4725. https://doi.org/10.1242/jeb.01950

Lichtwark, G. A., & Wilson, A. M. (2006). Interactions between the human gastrocnemius muscle and the Achilles tendon during incline, level and decline locomotion. *Journal of Experimental Biology*, *209*(21), 4379–4388. https://doi.org/10.1242/jeb.02434

McGinnis, P. M. (2013). *Biomechanics of sport and exercise* (3rd ed.). Human Kinetics.

Monte, A., Baltzopoulos, V., Maganaris, C. N., & Zamparo, P. (2020). Gastrocnemius Medialis and Vastus Lateralis in vivo muscle-tendon behavior during running at increasing speeds. *Scandinavian Journal of Medicine & Science in Sports*, *Online Version of Record before inclusion in an issue*. https://doi.org/10.1111/sms.13662

Nilsson, J., Tveit, P., & Eikrehagen, O. (2004). Cross-Country Skiing. *Sports Biomechanics*, *3*(1), 85–108. https://doi.org/10.1080/14763140408522832

Özgünen, K. T., Çelik, U., & Kurdak, S. S. (2010). Determination of an Optimal Threshold Value for Muscle Activity Detection in EMG Analysis. *Journal of Sports Science & Medicine*, *9*(4), 620–628.

Pellegrini, B., Zoppirolli, C., Bortolan, L., Zamparo, P., & Schena, F. (2014). Gait models and mechanical energy in three cross-country skiing techniques. *Journal of Experimental Biology*, *217*(21), 3910–3918. https://doi.org/10.1242/jeb.106740

Pellegrini, B., Zoppirolli, C., Stella, F., Bortolan, L., Holmberg, H.-C., & Schena, F. (2020). Biomechanical analysis of the "running" vs. "Conventional" diagonal stride uphill techniques as performed by elite cross-country skiers. *Journal of Sport and Health Science*. https://doi.org/10.1016/j.jshs.2020.04.011

Roberts, T. J., & Azizi, E. (2011). Flexible mechanisms: The diverse roles of biological springs in vertebrate movement. *Journal of Experimental Biology*, *214*(3), 353–361. https://doi.org/10.1242/jeb.038588

Roberts, T. J., & Belliveau, R. A. (2005). Sources of mechanical power for uphill running in humans. *Journal of Experimental Biology*, *208*(10), 1963–1970. https://doi.org/10.1242/jeb.01555

Robertson, D. G. E., Caldwell, G. E., Hamill, J., Kamen, G., & Whittlesey, S. N. (2014). *Research methods in biomechanics* (2nd ed.). Human Kinetics.

Rutherford, O. M., & Jones, D. A. (1992). Measurement of fibre pennation using ultrasound in the human quadriceps in vivo. *European Journal of Applied Physiology and Occupational Physiology*, *65*(5), 433–437. https://doi.org/10.1007/BF00243510

Smith, G. A. (2003). Biomechanics of cross country skiing. In H. Rusko, *Cross Country Skiing* (pp. 32–61). Blackwell Science Ltd.

Wang, J. H.-C. (2006). Mechanobiology of tendon. *Journal of Biomechanics*, *39*(9), 1563–1582. https://doi.org/10.1016/j.jbiomech.2005.05.011

Werkhausen, A., Cronin, N. J., Albracht, K., Bojsen-Møller, J., & Seynnes, O. R. (2019). Distinct muscle-tendon interaction during running at different speeds and in different

loading conditions. *Journal of Applied Physiology*, *127*(1), 246–253. https://doi.org/10.1152/japplphysiol.00710.2018

Whiting, W. C. (2008). *Biomechanics of musculoskeletal injury* (2nd ed.). Human Kinetics.

Zajac, F. E. (1989). Muscle and tendon: Properties, models, scaling, and application to biomechanics and motor control. *Critical Reviews in Biomedical Engineering*, *17*(4), 359–411.

Zatsiorsky, V. M., & Prilutsky, B. I. (2012). *Biomechanics of skeletal muscles*. Human Kinetics.

Zoppirolli, C., Hébert-Losier, K., Holmberg, H.-C., & Pellegrini, B. (2020). Biomechanical determinants of cross-country skiing performance: A systematic review. *Journal of Sports Sciences*, *38*(18), 2127–2148. https://doi.org/10.1080/02640414.2020.1775375

Corona statement

This thesis was written during the Covid-19 pandemic. Therefore, some measure was taking to ensure the safety of participants as well as fellow researchers. The data collection was done in September (2020) when the infection rate in Norway was relatively low. However, no more than one participant was allowed in the lab at the same time. The researcher used protective gear, including visor, face masks, latex gloves, and lab coats. All equipment used was disinfected between each test. Equipment that could not be disinfected was put in a quarantine at a minimum of 48 hours. Since campus was closed for most of the period, most of this thesis has been written from home, and most of the contact with supervisors has been over video calls.

Appendix 1: Approval letter from the Ethical Committee

Amelie Werkhausen

Institutt for fysisk prestasjonsevne   OSLO 13. juni 2020

Søknad 137 - 180620 – Når er diagonalgang effektivt? En studie av muskelsammentrekningsmodus i under ekstremitene og effekten av helning og hastighet

Vi viser til søknad, prosjektbeskrivelse, spørreskjema, informasjonsskriv, innsendt søknad til NSD og opplysninger om utvalget gitt i mail datert 4.6.2020

I henhold til retningslinjer for behandling av søknad til etisk komite for idrettsvitenskapelig forskning på mennesker, har leder av komiteen på fullmakt fra komiteen konkludert med følgende:

Vurdering:

Komiteen ser positivt på at man i studien legger opp til både menn og kvinner i utvalget, og at det tilstrebes en lik kjønnsfordeling i utvalget.

Vedtak

På bakgrunn av forelagte dokumentasjon finner komiteen at prosjektet er forsvarlig og at det kan gjennomføres innenfor rammene av anerkjente etiske forskningsetiske normer nedfelt i NIHs retningslinjer. Til vedtaket har komiteen v/leder lagt følgende forutsetning til grunn:

- *Vilkår fra NSD følges*
- *Det tilstrebes likt antall kjønn i utvalget*

Komiteen gjør oppmerksom på at vedtaket er avgrenset i tråd med fremlagte dokumentasjon. Dersom det gjøres vesentlige endringer i prosjektet som kan ha betydning for deltakernes helse og sikkerhet, skal dette legges fram for komiteen før eventuelle endringer kan iverksettes.

Med vennlig hilsen

Professor Sigmund Loland

Leder, Etisk komite, Norges idrettshøgskole

Vil du delta i forskningsprosjektet

Når er diagonalgang effektivt? En studie av muskelsammentrekningsmodus i underekstremitetene og effekten av helning og hastighet

Dette er et spørsmål til deg om å delta i et forskningsprosjekt hvor formålet er å studere muskel-sene dynamikken i diagonalgang ved ulike helninger på rulleskimølle.
I dette skrivet gir vi deg informasjon om målene for prosjektet og hva deltakelse vil innebære for deg.

Formål

I dette masterprosjektet ønsker vi å se på muskel-sene dynamikken under diagonalgang på rulleski ved ulike helninger på en rulleskimølle. Dette gjøres ved å kombinere ultralydavbildning av legg og lårmuskelen for å studere muskelfasciklenes lengde, sammentrekningshastighet og pennasjonsvinkel, med kinematiske målemetoder som 3D-bevegelsesregistering for å studere adferden til hele muskel-sene enheten. Kunnskapen dette gir kan øke vår forståelse av hvordan musklene skaper kraft i diagonalgang.

Spesifikt ønsker vi å se på om musklene i lår og legger trekker seg sammen konsentrisk for å skape kraft når vi går diagonalgang, eller om det er mulig og bevare mekanisk energi i sener og dermed tillate musklene å jobbe mer isometrisk, noe som i teorien kan bedre arbeidsøkonomien.

Hvem er ansvarlig for forskningsprosjektet?

Norges Idrettshøgskole (NIH) ved Institutt for fysisk prestasjonsevne (IFP) er ansvarlig for dette prosjektet.

Hvorfor får du spørsmål om å delta?

Du har fått spørsmål om å delta i dette forskningsprosjektet fordi vi ønsker å inkludere deltagere med følgende karakteristikker i vårt forskningsprosjekt.

Du du er aktiv eller tidligere aktiv skiløper som har; deltatt i enten Norges Cup (NC), Skandinavisk Cup eller Verdens cup (WC); minst 5 års erfaring med å gå på ski; ingen skader, eller andre lyter som hindrer deg i å gå diagonalgang; er fylt 18 år.

Hva innebærer det for deg å delta?

Hvis du velger å delta i prosjektet, innebærer det at du møter en dag i det biomekaniske laboratoriet ved NIH. Der vil du gjennomføre en testprotokoll på rulleskimølle som vil ta cirka 2-3 timer. Det bes om at du stiller i treningstøy, og helst en kort shorts. Ta med egne skisko og hansker. Hvis du ikke har NNN-kompatible støvler må du kontakte oss, så forsøker vi å finne en løsning.

For de kinematiske målingene vil det bli plassert refleksmarkører på anatomiske kjennetegn på kroppen din. Disse festes med dobbeltsidig tape. Muskelens aktivitetsnivå måles ved hjelp av overflateelektroder (elektromyografi, EMG) som tapes på musklene. To ultralydsonder vil festes over legg- og lårmuskelen for å måle fascikkellengde. Disse vil også festes til kroppen og holdes fast av et elastisk bånd. På grunn av EMG og ultralyd målingene må hunden barberes (og vaskes) der disse skal festes, **dette utføres på laboratoriet** av prosjektarbeiderne.
Staver / ski vil være utstyrt med sensorer som måler kraft mot underlaget.

Du vil, etter 10-15 minutters oppvarming på rulleski, gjennomføre fire korte drag hvor det gjøres dataopptak. Disse dragene vil vare i underkant av et minutt. Belastningen vil være nær konkurransefart, men grunnet den korte varigheten vil belastningen totalt sett ikke være svært anstrengende. Spesifikt skal du gå 2 drag i bratt motbakke (10°) på 2.50 m/s. På det ene draget skal du gå diagonalgang med glidfase, på det andre skal du gjennomføre «løpende» diagonalgang. Deretter skal du gå 2 drag i slak motbakke (5°), et drag på 4.25 m/s og et drag på 2.50 m/s. Begge disse dragene utføres i diagonalgang med glidfase.

Det er frivillig å delta

Det er frivillig å delta i prosjektet. Hvis du velger å delta, kan du når som helst trekke samtykket tilbake uten å oppgi noen grunn. Alle dine personopplysninger vil da bli slettet. Det vil ikke ha noen negative konsekvenser for deg hvis du ikke vil delta eller senere velger å trekke deg.

Ditt personvern – hvordan vi oppbevarer og bruker dine opplysninger

Vi vil bare bruke opplysningene om deg til formålene vi har fortalt om i dette skrivet. Vi behandler opplysningene konfidensielt og i samsvar med personvernregelverket.

- Det er kun personer i prosjektgruppen som skal ha tilgang til dine personopplysninger.
- Navnet og kontaktopplysningene dine vil bli erstattet med en kodenøkkel som lagres på en egen navneliste avskilt fra øvrige forskningsdata.
- Alt data materiale vil være passord beskyttet.
- Dine personopplysninger vil ikke bli send via email eller over internett.

Du vil ikke kunne gjenkjennes i en eventuell publikasjon

Hva skjer med opplysningene dine når vi avslutter forskningsprosjektet?

Prosjektet skal etter planen avsluttes 01.07.21. Vi er pliktet til a oppbevare data og separat navneliste i 5 år etter sluttdato for etterprøvbarhet og kontroll av resultatene. Etter dette, altså 01.06.26, vil all data i prosjektet slettes.

Dine rettigheter

Så lenge du kan identifiseres i datamaterialet, har du rett til:
- innsyn i hvilke personopplysninger som er registrert om deg, og å få utlevert en kopi av opplysningene,
- å få rettet personopplysninger om deg,
- å få slettet personopplysninger om deg, og
- å sende klage til Datatilsynet om behandlingen av dine personopplysninger.

Hva gir oss rett til å behandle personopplysninger om deg?

Vi behandler opplysninger om deg basert på ditt samtykke.

På oppdrag fra NIH har NSD – Norsk senter for forskningsdata AS vurdert at behandlingen av personopplysninger i dette prosjektet er i samsvar med personvernregelverket.

Hvor kan jeg finne ut mer?

Hvis du har spørsmål til studien, eller ønsker å benytte deg av dine rettigheter, ta kontakt med:

- Amelie Werkhausen / Øyvind Gløersen (prosjektansvarlig)
  (amelie.werkhausen@nih.no / o.n.gloersen@nih.no)
- Anders Lundervold (masterstudent)
  (alundervold@gmail.com / 90289919)
- Vårt personvernombud: Rolf Haavik (rolf.haavik@habberstad.no)

Hvis du har spørsmål knyttet til NSD sin vurdering av prosjektet, kan du ta kontakt med:

- NSD – Norsk senter for forskningsdata AS på epost (personverntjenester@nsd.no) eller på telefon: 55 58 21 17.

Samtykkeerklæring

Jeg har mottatt og forstått informasjon om prosjektet *«Når er diagonalgang effektivt? En studie av sammentrekningsmodus i underekstremitetene»*, og har fått anledning til å stille spørsmål. Jeg samtykker til:

☐ å delta i *testene på rulleskimølle, hvor det samles inn ultralyd bilder, kinematiske data, EMG signaler, og kraftmåling fra ski og staver,*

Jeg samtykker til at mine opplysninger behandles frem til prosjektet er avsluttet og lagres i 5 år etter prosjektslutt.

(Signert av prosjektdeltaker, dato)