Muscle fascicle behaviour during stair negotiation

Marcel Spanjaard

The research presented in this thesis was carried out at the Research Institute MOVE, Faculty of Human Movement Sciences, VU University, Amsterdam, The Netherlands in collaboration with the Institute for Biomedical Research into Human Movement and Health, Department of Exercise and Sport Science, Manchester Metropolitan University, Cheshire, United Kingdom.

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GASTROCNEMIUS MUSCLE FASCICLE BEHAVIOUR DURING STAIR NEGOTIATION

MARCEL SPANJAARD

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Institute for Biomedical Research into Human Movement and Health,
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Spanjaard M, Reeves ND, van Dieen JH, Baltzopoulos V, Maganaris CN. (2007) Effect of gait velocity on human gastrocnemius fascicle behaviour during stair ascent. *Proceedings of the XIIth Annual Congress of the European College of Sport Science*, Jyväskylä, Finland.

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Reeves ND, Spanjaard M, Mohagheghi AA, Baltzopoulos V, Maganaris CN. (2007) Older adults employ alternative strategies to operate within their maximum capabilities when ascending stairs. *J Electromyogr Kinesiol*, doi:10.1016/j.jelekin.2007.09.009

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SUMMARY

The present thesis investigated the gastrocnemius medialis (GM) muscle fascicle behaviour during stair negotiation to understand how the contractile behaviour of this muscle in this task is influenced by the series elastic component in conditions of different demands. The influence of physiologically meaningful variations in the task demands were investigated by manipulating gait velocity, step-height and body mass. Participants walked up and down a four-step staircase, during which kinetic data were acquired. Real-time ultrasound scanning was used to determine GM muscle fascicle length changes, while musculotendon complex (MTC) length changes were estimated from ankle and knee joint kinematics. During the lift-off phase of stair ascent, the GM muscle fascicles contracted near-isometrically, providing force, while the MTC shortened due to tendon recoil. The ankle joint moment peaked in this phase, also providing a positive joint power peak. During the touch-down phase of stair descent, there was a negative peak in the ankle joint power (and a high moment) due to MTC lengthening (ankle dorsiflexion). In contrast, however, the GM muscle fascicles shortened in this phase, performing positive work. In general, when the task demands increased (by increasing gait velocity or step-height), the GM muscle fascicles shortened more (even when the MTC lengthened during stair descent), mostly coinciding with an increased ankle joint moment. However, due to a change in strategy, increased body mass did not result in an increased GM muscle fascicle shortening. The main finding from this work is that the GM muscle fascicle behaviour does not parallel, and hence it cannot be predicted from, the MTC behaviour during stair negotiation. Further, although in general increasing the demands of stair negotiation lead to increased joint moments and more fascicle shortening, the response of the musculoskeletal system to altered demands is not predictable and seems to be specific to the method by which the demands are modified.

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Chapter 1

General Introduction

Structure and function of muscles in vivo

Voluntary human movement can only be generated by contracting muscles. The word “muscle” is derived from the Latin word “musculus”, which means “little mouse”. This name reflects the general form of muscle: a muscle belly attached to a tendon (the little mouse and its tail, respectively). More accurately described, the muscle and its tendon make up the musculotendon complex (MTC). The muscle consists of muscle fibres, which will attempt to shorten whenever there is an input signal from the alpha motor neurons in the spinal cord, which are in turn controlled by the motor cortex and lower level reflex loops. The activated muscle fibres pull on the tendon, and, in turn, the tendon pulls on the bone to which it is attached. If the pulling force is high enough relative to other forces working on the bone, this will result in a change in joint angle. Some of the muscles crossing a joint will pull on the bone in the same direction (agonists), while other muscles will pull the same bone in the opposite direction (antagonists). The bone will only rotate in one direction (if any) and therefore the antagonists will lengthen on agonist shortening or vice versa. If all the muscles contributing to a movement are activated in a coherent manner, then the movement of bones around joints will result in functional movement.

A muscle fibre is a long and thin cell. Its length can range from several millimetres up to 12 cm, while its diameter ranges from 10 to 80 micrometers [Wilmore and Costill, 1999]. Bundles of these muscle fibres, in series and in parallel, are wrapped in a connective tissue sheath (perimysium), these bundles are called fasciculi or muscle fascicles and, in some muscles, can be seen by the naked eye (Figure 1.1). Hundreds of these muscle fascicles make up the muscle belly, the contracting part of the musculotendon complex.

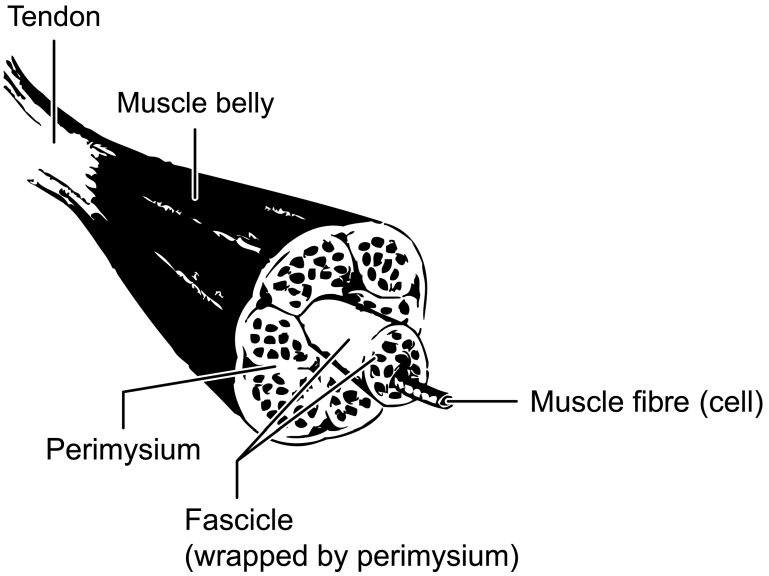

Figure 1.1. Drawing of a muscle belly in which the muscle fibres are wrapped by the perimysium. These bundles of muscle fibres are called “muscle fascicles”.

Muscle architecture, the arrangement of muscle fascicles and tendon, varies considerably between muscles. Architectural differences also imply functional differences, e.g. muscles with long fascicles and relatively short tendons can do a large amount of work with just little tendon stretch. This architecture is typical for the larger muscles that are found in the proximal segments of limbs (e.g. the hamstring muscles). Muscles with relatively short fascicles and long tendons facilitate energy storage and release during movements such as walking and running and are mainly found in the distal segments of limbs (e.g. the gastrocnemius muscles) [Alexander and Ker, 1990]. In some muscles, the muscle fascicles are obliquely attached to an aponeurosis (tendon sheet). The pennation angle is defined as the angle made by fascicles and the line of pull of muscle [Huijing, 1985; Kawakami et al., 1998]. Pennation allows more fascicles to be attached to a given length or area of aponeurosis.

When muscle fibres contract and the muscle fascicles shorten, the tendon and its aponeurosis are pulled and stretched. Since the tendon-aponeurosis complex, as an elastic element in series, might take up a large part of any length change along the entire MTC, the muscle fascicle length change is not necessarily similar to the MTC length change, which can be calculated from the changes in joint angles [Grieve et al., 1978; Hawkins and Hull, 1990; Menegaldo et al., 2004]. Consider, for example, the simplest scenario of the so-called "isometric" contraction. The constancy in joint configuration on a dynamometer ensures that the MTC will remain isometric during the contraction. However, the muscle fascicles will shorten as the muscle force develops, pulling on and stretching the in-series tendinous tissue until maximum contractile force is produced (Figures 1.2 and Figure 1.3) [Maganaris and Paul, 1999; Narici et al., 1996]. The degree of tendinous stretching during the development of contraction will determine the length of

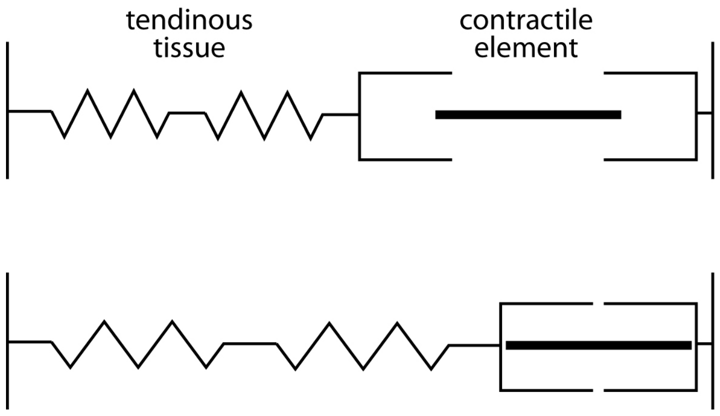

Figure 1.2. Schematic representation of a musculotendon complex (MTC) as a sarcomere-spring model at rest (top), and during an "isometric" contraction (bottom). The MTC length is constant while the lengthening of the tendinous tissue (the tendon and its aponeurosis) during contraction is taken up by the contractile element.

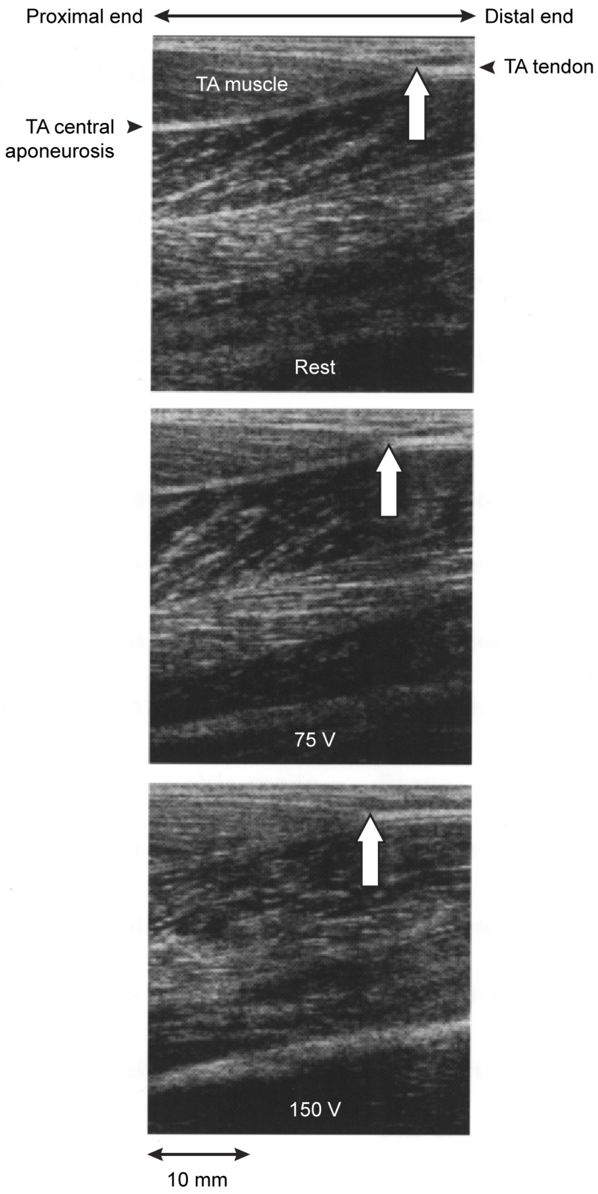

Figure 1.3. Sonographs over the musculotendinous junction of the tibialis anterior (TA) muscle at rest (top), and electrically stimulated at 75 V (middle) and 150 V (bottom), with the ankle joint (and hence the MTC length) fixed. The displacement of the TA tendon origin (white arrow) in the transition from rest to increasing intensity contractions shows the shortening of the muscle and elongation of tendon, consistent with the model shown in Figure 1.2. [Figure adapted from Maganaris and Paul [Maganaris and Paul, 1999]]

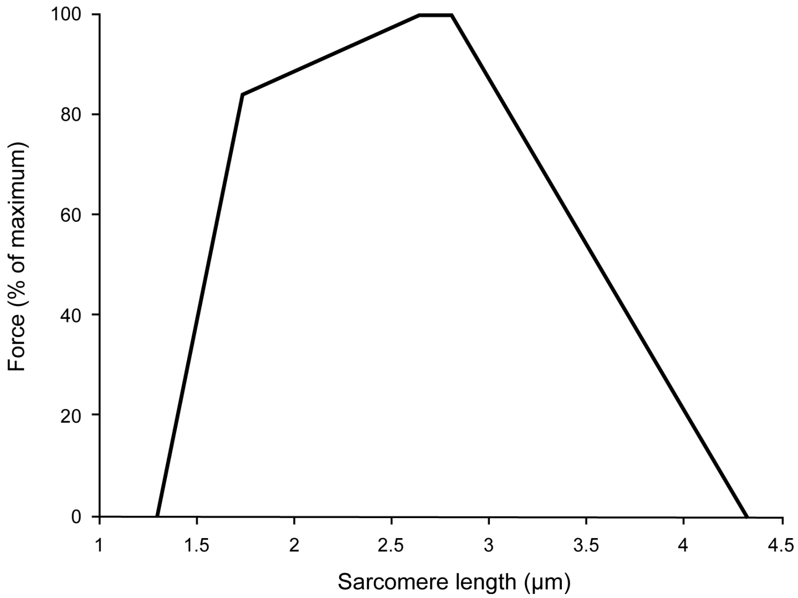

Figure 1.4. Theoretical length-tension relationship for a single human sarcomere, predicted by the cross-bridge model of contraction [Gordon et al., 1966] using human myofilament length data [Walker and Schrodt, 1974].

the muscle fascicles in the steady-state phase of the contraction, which, in turn, will determine the length of the sarcomeres during contraction and hence their force-generating capability based on the force-length relationship (Figure 1.4). During dynamic - whole body - movements, the joint configuration and hence the MTC length of the muscles involved in the movement varies, and so does the level of activation in the muscles and the stiffness of the series tendinous tissue [Herbert et al., 2002; Morse et al., 2008].

In addition, the speed of contraction may vary, which will impact on the force developed by the muscle fibres, and thus that applied to the tendon according to the force-velocity relationship. Figure 1.5 shows the relationship between force and contraction velocity in an idealised muscle [Jones et al., 2004]. This shows that the faster the muscle fibres shorten, the less force they can produce, eventually reaching a velocity at which force can no longer be sustained. When the activated muscle fascicles lengthen (eccentric muscle fibre action), the force produced increases; in this case the faster the muscle stretch, the higher the force produced, up to a certain maximum [Jones et al., 2004]. An increasing series compliance will change the force-velocity relationship of the whole MTC. If the movement starts from an isometric state the overall shortening or lengthening of the MTC will be shared between the tendon and the muscle fibres in proportion to their compliances, the net result being that the muscle fibres will be moving more slowly than the whole MTC. During shortening at the same speed of the MTC, a muscle with a more compliant tendon will generate more force than one with a lesser compliance. During lengthening, however, the muscle with the more compliant tendon will generate somewhat less force. The length-tension relationship of tendon is not linear, however, so that during stretch the tendon will be relatively stiff compared to the muscle fibres and thus differences in tendon compliance may make little difference.

The type and velocity of contraction will also influence the energetic cost of contraction. The amount of energy required depends on the rate of attachment and detachment of cross-bridges (which cause the contraction of muscle). During lengthening contractions, the cross-bridges attach and provide stiffness to the muscle, but there is hardly any energy required for their detachment, since this is caused by the external force applied. For isometric and shortening contractions on the other hand, the cross-bridges will have to attach, contract and detach, which results in higher energy requirements. When shortening velocity increases, the rate of detachment increases, resulting in higher energy expenditure [Jones et al., 2004]. Since the tendon influences the rate of shortening of the muscle fascicles, it also influences the energy expenditure. As explained above, the fascicles are only responsible for part of the shortening of the whole MTC. Therefore, the rise in energetic expenditure with higher contraction velocities is reduced by the presence of compliant tendinous tissue.

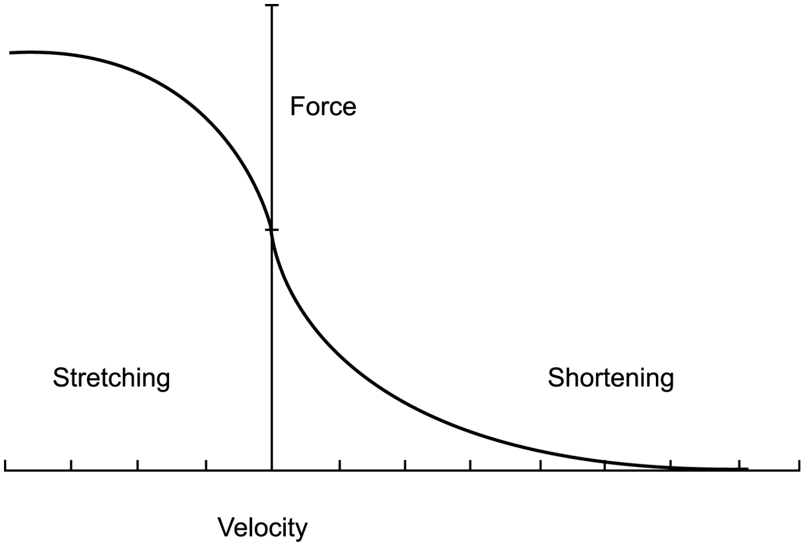

Figure 1.5. Schematic representation of the muscle force-velocity relationship.

The factors described above, which affect the interaction between muscle fascicles and series tendinous tissue, make the estimation of contractile behaviour from MTC behaviour unreliable. The outcome of the interaction between muscle and tendon at given point in time during movement depends on the "state" of both these two functional structures. Being able to differentiate the two behaviours can provide insight not only into how the musculoskeletal system functions, but also into how the central nervous system coordinates movements.

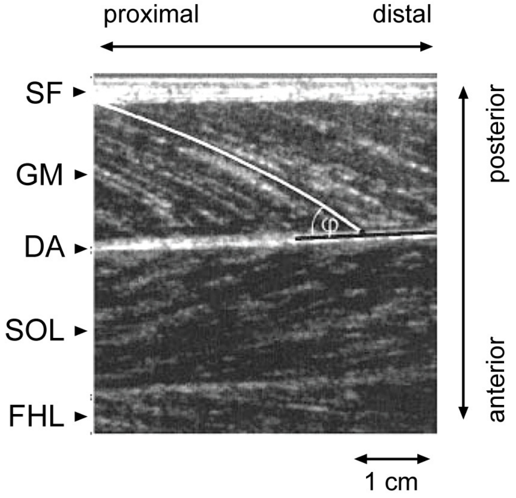

Figure 1.6. Typical sonograph of the gastrocnemius medialis muscle. The fascicular trajectory between the two aponeurosis, as well as the pennation angle (φ) are highlighted in white. SF, subcutaneous fat and superficial aponeurosis; GM, gastrocnemius medialis muscle; DA, deep aponeurosis; SOL, soleus muscle; FHL, flexor hallucis longus muscle. The ultrasound probe is on the skin above the subcutaneous fat layer.

A relatively new way of measuring muscle fascicle length in vivo – non-invasively – is provided by the use of ultrasonography (Figure 1.6). An ultrasound probe (Figure 1.7) is placed onto the skin and sends out beams of ultrahigh frequency (typically in the range ~ 5-15 MHz) sound waves. The sound waves are reflected by tissue to form echoes of the original waves. From the echoes that the ultrasound probe detects an image can be constructed. The direction and intensity of the echoes and the time delay from sending to receiving are used to construct the ultrasound image. From a variety of ultrasound imaging techniques, linear B-mode ultrasonography is used to visualise muscle fascicles. With this ultrasound technique, a 2-D image is made in the plane in which the probe is positioned and length measurements can directly be related to the distances in the tissue itself. This immediately illustrates the importance of correct placement of the ultrasound probe: the plane that is imaged must be the plane in which the muscle fascicles operate. Klimstra et al. [2007] showed that when the probe is not placed correctly, the fascicles in the constructed image may appear longer or shorter than they are because of a

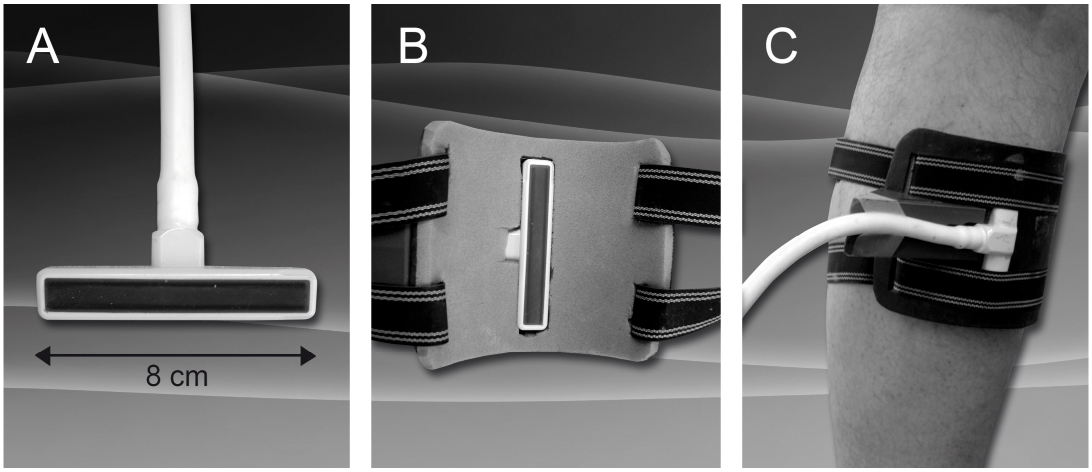

Figure 1.7. A linear 7.5 MHz probe (A) with 60 mm field of view (8 cm long). In the present work, a custom-built fixation device made of Velcro straps and a plastic cast molded to fit the general contour of the calf (B) was used to secure the probe around the left lower leg, in the mid-sagittal plane of the GM muscle (C).

projection error. When muscle fascicles on the ultrasound image appear to have little gaps in them, this may be due to a projection error (the fascicle with little gaps actually consist of multiple fascicles lying side by side that are viewed as one). In a good ultrasound image for measuring fascicle length, the fascicles appear as single long entities between the two aponeuroses. Note that the white lines on the ultrasound image are in fact fragments of connective tissue (perimysium), which contain the fascicles (Figure 1.6).

Since ultrasonography was first introduced for the study of muscle architecture in vivo [Kawakami et al., 1993; Narici et al., 1996; Rutherford and Jones, 1992], several experiments have been conducted to determine changes in muscle fascicle length in different conditions and tasks [Antonios and Adds, 2008; Chino et al., 2008; Fukashiro et al., 2006; Fukunaga et al., 2001; Ishikawa et al., 2005a; Ishikawa et al., 2005b; Ishikawa et al., 2007; Kawakami et al., 1998; Kawakami et al., 2002; Kawakami and Fukunaga, 2006; Kawakami et al., 2000; Klimstra et al., 2007; Kubo et al., 1999; Kubo et al., 2000; Kurokawa et al., 2001; Kurokawa et al., 2003; Lichtwark and Wilson, 2005; Lichtwark and Wilson, 2006; Mademli and Arampatzis, 2008; Maganaris et al., 1998; Maganaris, 2003; Maganaris et al., 2006; Narici et al., 2003; Reeves and Narici, 2003]. A common finding is that MTC length changes do not correspond one-to-one with fascicle length changes. This in not a surprising finding, since, as discussed above, the length of the tendon-aponeurosis complex, the elastic element in series with the fascicles, changes under the influence of force generation. Nevertheless, the behaviour of muscle fascicles (fascicle length and fascicle shortening velocity) is often estimated from joint kinematics, ignoring the compliance of the tendon.

In recent years, research into muscular behaviour in vivo has been focused on the interaction fascicles-tendon, especially in movements which require mainly propulsive forces, such as walking, running and jumping [Fukunaga et al., 2001; Hof et al., 2002; Ishikawa et al., 2005a; Ishikawa et al., 2007; Kawakami et al., 2002; Kurokawa et al., 2001; Kurokawa et al., 2003; Lichtwark and Wilson, 2005]. Most of the studies suggest that muscles producing the propulsion behave highly effectively: muscle fascicles are presumed to produce high levels of force because they are often seen to contract near-isometrically [Fukunaga et al., 2001; Kawakami and Fukunaga, 2006; Lichtwark and Wilson, 2006; Roberts, 2002], whereas the tendon stores (during stretch) and releases energy (during recoil) and works as a power amplifier [Alexander and Bennet-Clark, 1977; Fukunaga et al., 2001; Kawakami and Fukunaga, 2006; Roberts, 2002]. This process of MTC stretch and recoil is known as the stretch-shortening cycle, and can increase performance because of its energetic efficiency [Komi, 1992]. However, movement also requires muscles to function as brakes (energy absorbing elements) by contracting eccentrically [Lindstedt et al., 2001]. An active lengthening occurs when the external force applied to the muscle exceeds that produced by the muscle, thereby absorbing energy. The fundamental properties of an eccentric contraction imply a low energy cost and a high force production [Bigland-Ritchie and Woods, 1976; Lindstedt et al., 2001].

Stair negotiation

A comparison of locomotion on stairs with the most basic form of locomotion, level gait, is insightful for realising the demanding nature of stair negotiation. Level walking has frequently been studied to understand the energetic requirements and mechanics of major muscle groups involved [Collins et al., 2005; Donelan et al., 2002a; Donelan et al., 2002b; Fukunaga et al., 2001; Kuo, 2002; Winter, 1987]. The exchange of kinetic and potential energy in each step makes it an energetically efficient movement, in which the interaction between muscle fascicles (contracting isometrically during the stance phase) and tendon (storing and releasing energy) at the ankle play an important role [Fukunaga et al., 2001]. The muscles (ankle plantarflexors) do not need to perform a large amount of positive or negative work, since in steady state level walking the total net energy (kinetic plus potential energy) remains constant [Kuo, 2007].

When walking up or down, on stairs or slopes, the total amount of energy changes. This is reflected by changes in kinematics and kinetics; both the ranges of motion in lower extremity joints and the joint moments increase when walking on slopes or stairs compared to level walking [Devita et al., 2007; Lichtwark and Wilson, 2006; Riener et al., 2002]. The height difference requires the musculoskeletal system to perform more positive or negative work. Studies that looked at plantarflexor muscle fascicle behaviour in running turkey's (using sonomicrometry) have shown: 1 – Isometric contraction during level running, where the storage and release of energy in the tendon is sufficient to maintain speed and the fascicles only need to supply force without shortening; 2 – Concentric contraction during up-slope running, where extra shortening (positive mechanical work) is required to lift the body upwards; 3 – Eccentric contraction during down-slope running, where energy needs to be absorbed (negative mechanical work) [Gabaldon et al., 2004; Roberts et al., 1997; Roberts, 2002]. However, in humans walking up or down a slope, the ankle plantarflexors behave in the same way as in level walking: they contract isometrically during stance and shorten somewhat during the lift-off phase [Fukunaga et al., 2001; Lichtwark and Wilson, 2006].

In the above mentioned experiment by Lichtwark and Wilson [Lichtwark and Wilson, 2006], subjects negotiated slopes of 10%. In contrast, stair walking involves inclines of 50 to 60% in public places and it can go up to inclines of about 90% [Roys, 2001]. Stair ascent aims at elevating the human body, so muscle fascicles must generate energy in order to accomplish the required increase in joint moments and joint work [Devita et al., 2007; Riener et al., 2002]. Stair descent aims at lowering the human body, going from a state of higher potential energy to a state of lower potential energy without acquiring additional kinetic energy (increasing in gait velocity), which means that muscles will have to work as brakes, providing negative work – absorbing energy. Thus, joint moments will oppose the ongoing joint rotations, and are known to increase in magnitude when compared to level walking [Devita et al., 2007; Riener et al., 2002]. To gain knowledge about how muscles generate and absorb the required energy during stair ascent and descent, respectively, muscle fascicle behaviour during stair negotiation is investigated in the present thesis and related to joint kinematics and kinetics. Accurate information on muscle fascicle length and its changes is of vital importance for gaining insight into mechanical and energetic aspects of locomotion [Taylor and Heglund, 1982] and motor control.

It is known that the joints of the lower extremity – in particular the knee and ankle joints – are crucial in stair negotiation [McFadyen and Winter, 1988; Protopapadaki et al., 2007; Riener et al., 2002] and that the basic mechanical patterns observed are reproducible [McFadyen and Winter, 1988; Yu et al., 1997]. In the previous paragraph, it was explained that when negotiating stairs, the muscles need to perform an adequate amount of positive or negative work to move the body upwards or control its downward movement. Although muscles are the real actuators of movement, previous studies into stair negotiation mainly reported kinematic and kinetic data of the joints involved. Compared to level walking, the ground reaction forces and joint moments are higher during stair walking [Andriacchi et al., 1980; McFadyen and Winter, 1988; Stacoff et al., 2005], especially in the lift-off phase during stair ascent and the touch-down phase during stair descent [McFadyen and Winter, 1988; Protopapadaki et al., 2007; Riener et al., 2002]. The higher joint moments in stair walking compared to level walking are associated with an increase in joint range of motion, which also increases the difficulty of the task.

Furthermore, joint moment analysis has shown that the knee joint can reach moment values of about 50% of maximal voluntary contraction (MVC) in stair ascent and 30% of MVC in stair descent, whereas the ankle joint reaches moment values of about 85% of MVC in stair ascent and 75% of MVC in stair descent [Reeves et al., 2007; Reeves et al., 2008]. These values indicate that the normal daily activity of stair negotiation can be a strenuous task, especially for frail people such as the elderly, and this is further exemplified by the high incidents of falls and accidents on stairs [Cavanagh et al., 1997; Hemenway et al., 1994; Lord and Dayhew, 2001; Moseley et al., 2003; Riener et al., 2002; Simoneau et al., 1991; Stacoff et al., 2005; Svanstrom, 1974; Tinetti, 2001]. Therefore, it is not surprising that stair walking is often used in clinical settings for evaluating rehabilitation outcomes [Bean et al., 2007; Sohng et al., 2003], or as a training activity [de Vreede et al., 2004; de Vreede et al., 2007] aimed at improving the quality of life.

Riener et al. [2002] investigated the kinematics and kinetics of stair negotiation and changed the inclination of the staircase. Joint angles and moments showed a relatively low but significant increase with inclination, which is in line with other studies [Livingston et al., 1991; Riener et al., 2002; Stacoff et al., 2005]. However, a large influence of inclination was observed in joint powers. The authors concluded that this could be related to the varying amount of potential energy that has to be produced (during ascent) or absorbed (during descent) on stairs of different dimensions.

The knee and ankle joints make a large contribution in terms of joint moment and power during stair negotiation. The joint moments and powers are generated predominantly by the knee extensor and ankle plantarflexor muscles [McFadyen and Winter, 1988; Townsend et al., 1978]. This is best shown in a study by McFadyen and Winter [1988], in which the electromyographical (EMG) activity of the main leg muscles was related to the calculated joint moments. During ascent, the knee extensor muscles (EMG was sampled from the rectus femoris and vastus lateralis muscles) and the hip extensor muscles (EMG was sampled from the gluteus maximus and medius muscles) were found to be active in the stance phase, in particular during the first part of stance, thereby pulling the body upwards with knee and hip extension. The ankle plantarflexors (EMG was sampled from the GM and soleus muscles) were found to be active during the second part of the stance phase, just before lift-off, pushing the body upwards (while at around the same time the hip and knee joint of the contra-lateral leg were pulling the body upwards) [McFadyen and Winter, 1988; Riener et al., 2002]. In the swing phase, the knee flexor muscles (EMG was sampled from the semitendinosus muscles) and the ankle dorsiflexor muscles (EMG was sampled from the tibialis anterior muscle) were found to be active, to allow clearance of the steps in the staircase.

During descent, the knee extensor muscles (EMG was sampled from the rectus femoris and vastus lateralis muscles) were found to be active throughout the stance phase, producing an eccentric knee extension moment. The hip extensors (EMG was sampled from the gluteus maximus and medius muscles) were found to be hardly active and therefore barely produced any joint moment. The ankle plantarflexor muscles were active in the stance phase, the GM muscle showed a significant EMG burst during touch-down, while the soleus muscle was activated moderately throughout the stance phase, both contributing to the ankle plantarflexion moment, which was eccentric at touch-down. In some subjects, the GM muscle was found to be activated in late swing, suggesting that some impact preparation was made [Craik et al., 1982; Freedman et al., 1976; Greenwood and Hopkins, 1976; McFadyen and Winter, 1988; Townsend et al., 1978]. During the swing phase, the knee flexor muscles (EMG was sampled from the semitendinosus muscle) and hip extensor muscles (EMG was sampled from the rectus femoris muscle) were found to be active [McFadyen and Winter, 1988], again to provide clearance.

Absolute joint power produced at the ankle joint is slightly higher than at the knee joint during stair negotiation [Duncan et al., 1997; Riener et al., 2002]. We have previously demonstrated that the ankle joint is loaded relatively more than the knee joint, which suggests that the ankle joint plays a crucial and possibly limiting role in stair negotiation [Reeves et al., 2007; Reeves et al., 2008]. Furthermore, ankle plantarflexion moment generation just before lift-off during stair ascent, and just after touch-down during stair descent, contributes substantially to the energy production and absorption for stair ascent and descent, respectively [Devita et al., 2007; Riener et al., 2002]. Therefore, fascicle behaviour of the triceps surae muscle (the major plantarflexor muscle group) appears vital for the generation of positive and negative work during stair negotiation and is therefore investigated in this thesis.

The triceps surae muscle consists of the GM muscle, the gastrocnemius lateralis (GL) muscle and the soleus muscle. The soleus muscle lies underneath the two gastrocnemii and is uni-articular, it only crosses the ankle joint. The two gastrocnemii muscle heads are bi-articular, they cross both the ankle and knee joints. An important difference between the gastrocnemii and the soleus muscles is the muscle fibre type. The human GM muscle is composed of 36-50% fast-twitch fibres, while the soleus muscle is composed out of 20-30% fast-twitch fibres [Saltin and Gollnick, 1983]. This means that the GM is a “faster” muscle, and is therefore more likely to have a relatively larger contribution in faster, more powerful and more strenuous movements [Carpentier et al., 1996; Duchateau et al., 1986; Duysens et al., 1991; Nardone and Schieppati, 1988; Tamaki et al., 1997; Vandervoort and McComas, 1983]. Furthermore, from dynamometry-based studies it is known that the soleus muscle is already fully activated at ~70% of MVC [Maganaris et al., 2006] and that further modulation of the joint moment is accomplished by gastrocnemii activation. This also seems to hold for stair negotiation: EMG data indicates that the soleus muscle has a relatively low level of activation throughout the stride cycle, whereas the GM muscle shows phasic and higher bursts of activation [McFadyen and Winter, 1988]. Since the ankle joint moment reaches values of about 75-85% of MVC during stair negotiation, it is anticipated that the contributions of the gastrocnemii are considerable during this task. It is known that the volume of the GM muscle is almost twice as much as that of the GL muscle, furthermore, the physiological cross sectional area is about three times larger for the GM muscle [Fukunaga et al., 1992; Fukunaga et al., 1996]. This indicates that the GM muscle contributes more to power and force production than the GL muscle. Therefore, the present thesis focuses on the fascicle behaviour of the GM muscle.

The demands placed on the musculoskeletal system during stair negotiation are quite substantial as discussed above, not in the least part because of the large amount of positive or negative work that has to be performed. The amount of work that has to be performed is predominantly determined by the dimensions of the stairs and the body mass of the person (since these influence the total amount of work performed in a single stride). Stair walking velocity defines how fast this work has to be performed, thereby influencing the power requirements of the task. It is anticipated that the amount of GM muscle fascicle shortening or lengthening is related to the amount of (positive or negative) work performed. Increasing the demands of stair negotiation is an interesting experimental manipulation since this will cause the – already substantial – joint moments to increase even more, approaching further the maximal capabilities of the musculoskeletal system. In the present thesis, the demands of stair negotiation will be increased by increasing gait velocity, step-height and body mass. These demands will influence the amount of positive or negative work that needs to be performed by the muscles (or the time it needs to be performed, in case of gait velocity). Besides being interesting experimental manipulations, the variations in gait velocity, step-height or body mass also are relevant to physiological situations and conditions encountered daily. Varying gait velocity is the main strategy adopted to negotiate everyday stepping tasks faster or slower. Step-height varies mainly with the location of the staircase [Roys, 2001], while the variation in total body mass represents a variation in the relative proportion of lean body mass.

It has been shown that an increase in step-height during stair negotiation requires an increase in ankle joint moment and power [Riener et al., 2002]. Increase in trunk mass during drop-landings has also shown to increase the ankle joint moment [Kulas et al., 2008]. The increase in ankle joint moment and power will be caused – for a large part – by an increased force and power production in the ankle plantarflexor MTCs. As discussed above, however, to understand the role played by the actuator of movement – the contractile element alone – it is necessary to differentiate the behaviour of the muscle fascicles from the behaviour of tendons and establish the effect on this interaction of the increasing-demand paradigms. It is hypothesized that the GM muscle fascicles will shorten more at higher demands during stair ascent, and lengthen more at higher demands during stair descent to satisfy the new requirements. This dependency on increased demands of muscle fascicle behaviour has already been shown in a study on running turkeys [Gabaldon et al., 2004]. In the latter mentioned study, sonomicrometry was used to identify muscle fascicle length changes of turkeys running on a slope which was increased in incline or decline. The running velocity was also increased, which resulted in muscle fascicles increasing the amount of shortening during uphill running and increasing the amount of lengthening during downhill running. Hence, increasing gait velocity in human stair ascent will probably cause an increase in GM muscle fascicle shortening, whereas it will probably cause an increase in GM muscle fascicle lengthening during stair descent.

Aim and outline

The general aim of this thesis was to gain insight into the interaction between muscle and tendon in a complex locomotor task and understand the mechanisms which modulate contractile behaviour and strategy when the task demand varies. To fulfil this aim, the GM muscle fascicle behaviour during stair negotiation and the effects on this behaviour of gait velocity, step-height and body mass were investigated.

Chapter 1 is the general introduction to the thesis and gives a brief overview of muscle mechanics and stair negotiation.

In **chapter 2**, GM muscle fascicle behaviour was measured during stair ascent and descent on a staircase of standard dimensions at a self selected gait velocity. The fascicular behaviour was related to kinematics and kinetics of the lower leg which were determined using motion capture and ground reaction forces.

Movement velocity is known to change fascicular behaviour; therefore, the influence of gait velocity on GM muscle fascicle behaviour during stair negotiation was investigated in **chapter 3**.

In the last 2 experimental chapters, task demands of stair ascent and descent were altered (altered step-height and body mass) and GM muscle fascicle behaviour was determined. In **chapter 4**, the influence of step-height and body mass on GM muscle fascicle behaviour during stair ascent was investigated. In **chapter 5**, the influence of step-height and body mass on GM muscle fascicle behaviour during stair descent was investigated.

In **chapter 6**, the main results are summarized and discussed, final conclusions are drawn, and future directions are suggested.

Chapter 2

Gastrocnemius muscle fascicle behaviour during stair negotiation

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Abstract

The aim of the present study was to establish the behaviour of human gastrocnemius medialis (GM) muscle fascicles during stair negotiation. Ten healthy male subjects performed normal stair ascent and descent at their own comfortable speed on a standard-dimension four-step staircase with embedded force platforms in each step. Kinematic, kinetic and electromyographical data of the lower limbs were collected. Real-time ultrasound scanning was used to determine GM muscle fascicle length changes. Musculotendon complex (MTC) length changes were estimated from ankle and knee joint kinematics. The GM muscle was mainly active during the push-off phase in stair ascent and the muscle fascicles contracted near-isometrically. The GM muscle was mainly active during the touch-down phase of stair descent where the MTC was lengthened, however, the GM muscle fascicles shortened by about 7 mm. These findings show that the behaviour and function of GM muscle fascicles in stair negotiation is different from that expected on the basis of length changes of the MTC as derived from joint kinematics.

Introduction

Stairs are regularly encountered in daily life and negotiating stairs is a more physically demanding task then level walking [Riener et al., 2002; Svanstrom, 1974]. The demands placed on the musculoskeletal system when ascending or descending stairs are not fully understood despite some research on stair negotiation [Andriacchi et al., 1980; Lark et al., 2003; Lark et al., 2004; McFadyen and Winter, 1988; Moseley et al., 2003; Riener et al., 2002; Stacoff et al., 2005; Tinetti, 2001; Townsend et al., 1978]. The ankle and knee extensor muscles are very important in stair negotiation, as they provide the moments and power needed to perform an ascent and to control a descent [Andriacchi et al., 1980; Lark et al., 2003; Lark et al., 2004; McFadyen and Winter, 1988; Moseley et al., 2003; Riener et al., 2002; Stacoff et al., 2005; Tinetti, 2001; Townsend et al., 1978]. In stair descent, the joint moment at the ankle can be as high as ~75% of a maximal voluntary contraction (MVC), whereas this is only ~30% at the knee [Reeves et al., 2008]. Also during stair ascent, the ankle joint appears to play a key role, with moments in ascent of comparable magnitude to descent [McFadyen and Winter, 1988].

At ankle plantarflexor moments higher than 70% of MVC, the soleus muscle is maximally activated [Maganaris et al., 2006], and thus only the gastrocnemius muscle allows further modulation of the plantarflexor moment. Hence, gastrocnemius muscle function may be crucial in stair negotiation. This is supported by the strong modulation of the electromyographic amplitude (EMG) of the gastrocnemius medialis (GM) muscle, compared to a much less pronounced modulation of the EMG amplitude of the soleus muscle in both ascent and descent [McFadyen and Winter, 1988]. McFadyen and Winter [1988] stated that the ankle extensor muscles act eccentrically in stair descent to dissipate a substantial amount of energy and act concentrically in ascent to produce a major part of the positive work required. However, because of the bi-articular nature of the gastrocnemius musculotendon complex (MTC), it is difficult to predict its behaviour. Furthermore, even when the kinematics of both the knee and ankle joints are taken into account to estimate gastrocnemius MTC length changes [Grieve et al., 1978; Menegaldo et al., 2004], this may not adequately reflect the behaviour of the contractile machinery. Because of elastic tendon structures and muscle pennation, the muscle fascicles do not necessarily undergo the same length changes as the whole MTC [Fukunaga et al., 2001; Kawakami et al., 2002; Lichtwark and Wilson, 2006].

Real-time ultrasonography allows reliable and non-invasive dimensional measurements of intact human fascicles [Kawakami et al., 1993; Maganaris et al., 1998; Narici et al., 1996]. Using this technique, it has been shown, that, in level walking, the GM muscle fascicle length stays relatively constant when the muscle is active during single support, while the tendon is being stretched to release elastic energy in the subsequent push-off phase [Fukunaga et al., 2001]. This behaviour is preserved when walking up and down slopes of 10% [Lichtwark and Wilson, 2006]. Thus across different modes of locomotion, this muscle acts more or less isometrically, which has been suggested to be energetically efficient [Fukunaga et al., 2001]. In the modes of locomotion studied heel landing is common, while in stair negotiation forefoot landing is commonly observed. This may imply that the previous findings on GM muscle behaviour do not generalize to stair negotiation, which might account in part for the demanding nature of this activity.

The purpose of this study was to establish the behaviour of the human GM muscle fascicles during stair negotiation. We hypothesized that GM muscle fascicles would remain at constant length while the muscle was active in both ascent and descent and we examined the relation between muscle fascicle and MTC length changes.

Methods

Subjects

Ten healthy young men [age: 24.9 ± 3.2 years, body mass: 79.9 ± 9.1 kg, height: 1.82 ± 0.06 m; mean ± standard deviation (SD), leg length (distance between the medial malleolus and spina iliaca anterior superior): 95.7 cm, ranging from 86 to 102.5 cm] volunteered to participate in the experiment. All volunteers gave their written consent to participate after approval was gained from the ethics committee of the Institute for Biophysical and Clinical Research into Human Movement at the Manchester Metropolitan University.

Measurements

Subjects walked up and down a custom-built steel staircase of four steps (Figure 2.1). The steps were independently mounted on the floor and their dimensions were 170 mm (riser) x 280 mm (tread) x 900 mm (width).

Kinetic data during stair ascent and descent were collected using four forceplates. Three force-plates (Kistler Z17068, 270 x 500 mm) with built-in amplifiers were embedded in the first three steps (from the ground), and one force plate (Kistler 9253A, 400 by 600 mm) with an external amplifier (Kistler 9865C) was embedded in the floor, in front of the staircase.

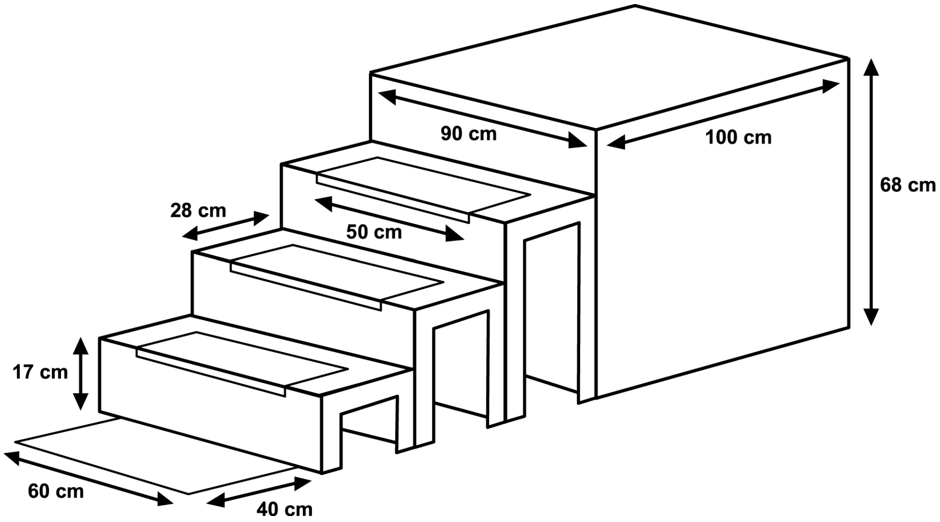

Figure 2.1. The custom-built steel staircase with step dimensions. Forceplates are embedded in the first 3 steps and in the floor in front of the staircase. The 4 steps are independently mounted in the floor.

Kinematic data were acquired using a 9-camera VICON 612 system (VICON motion systems Ltd., Oxford UK). Retro-reflective markers were placed on bony landmarks, directly on the skin, or on appropriate tight-fitting clothing using double-sided tape. In total, 34 markers were placed on the body according to the standard "plug-in-gait" model of the VICON system implemented in the Bodybuilder software module for 3-D segment modelling and calculation of upper and lower limb kinematics. Four markers were placed on the head: left and right temple, front of the head, and left and right on the back of the head (using a headband). Two markers were placed on the spinal column (C7 and T10), one in the centre of the right scapula, and two markers on the sternum (one cranial and one caudal). Three markers were placed on the pelvis, one on the sacrum and one on each spina iliaca anterior superior. Upper extremity markers were placed on the shoulder (acromion), elbow (brachial lateral epicondyle), two on the wrist (radial and ulnar directions using a wrist-bar) and one on the hand (second metacarpal). Lower extremity markers were placed on the lateral thigh (extended from the thigh using a wandmarker), knee (femoral lateral epicondylle), lateral side of the lower leg (extended using a wandmarker), ankle (lateral malleolus) and two on the foot (back of the heel and one on the second metatarsal bone). From the marker coordinates, knee and ankle joint angles were calculated. Knee joint angle was defined as the angle between the lower leg and a line through the thigh, so a straight leg corresponds to 0 deg knee joint angle. Positive angles indicated knee flexion. Ankle joint angle was defined as the angle of the foot with the lower leg, with 0 deg defined as the ankle at the neutral position. Positive angles indicated dorsiflexion and negative angles indicated plantarflexion.

The GM muscle behaviour was assessed in vivo from ultrasound scans recorded in real-time during the stair negotiation trials. For these measurements, a linear 7.5 MHz probe (UST-579T-7.5, Aloka SSD-5000, Tokyo, Japan) with 60 mm field of view was tightly secured around the left lower leg in the mid-sagittal plane of the GM muscle with a custom-built fixation device. The ultrasound scanning was synchronized with the kinematic and kinetic data using an external trigger.

The electrical activity of the GM muscle of the left leg was recorded using a Bagnoli EMG system (Delsys Inc. Boston, MA, USA). The recording electrodes were placed proximal to the ultrasound scanning probe in the mid-sagittal plane of the muscle. The sampling frequency of the EMG recordings was 2000 Hz.

To obtain an estimate of the magnitude of GM muscle EMG activity during stair negotiation relative to a plantarflexion MVC, we performed measurements in a sub-sample of 6 participants. EMG activity was recorded during stair negotiation and during MVC at 0 deg ankle joint angle. Participants performed five stair ascent and five stair descent trials on the previously described staircase and were then asked to perform

MVCs while lying prone on the bench of an isokinetic dynamometer (Cybex Norm, New York USA). For these measurements, EMG system, setup and placement of electrodes were maintained constant.

Protocol

Before the experiment, a number of anthropometric measurements were taken for each subject to scale the generic human plug-in-gait model in the VICON software (Oxford Metrics Inc.). Subsequently, the markers and EMG electrodes were positioned and data collection was initiated.

Subjects performed three trials of stair descent and three trials of stair ascent at a self-selected pace, barefoot, in a step-over-step fashion. Subjects stood still on top of the platform (stair descent), or on the ground just in front of the ground force-plate (stair ascent), and started every trial with their right foot. The trial ended when the subject was on the top platform, or on the ground off the force-plate, with the two feet together. All six trials were recorded, but only the last trial in each direction was analyzed to ensure that the subjects were familiarized with the tests.

Data analysis

The step between the first touchdown point of the left foot and the second touchdown point of the left foot (two steps above/below) was considered a steady-state step [Andriacchi et al., 1980]. From marker positions and forceplate data, VICON software was used to calculate kinematics and kinetics in 3-D using the plug-in-gait model. The kinematic and kinetic data of the ankle and knee, from the steady-state step were transported from ‘VICON workstation’ software to Matlab (The Mathworks, Inc, Natick, MA).

The GM muscle fascicle lengths were measured from the sonographs recorded. On each ultrasound frame during the steady-state step, three points were located manually using Matlab: one point at the end of the fascicle on the superficial aponeurosis, a second point at the end of the same fascicle on the deep aponeurosis, and a third point along the deep aponeurosis to allow calculation of the pennation angle. From these three points, the fascicle length and pennation angle were calculated. The muscle fascicle length was measured using the assumption that the fascicular trajectory was linear. The pennation angle was the angle that the fascicle made with the deep aponeurosis. The fascicle length measured in a standing position was the reference length for each subject. To account for individual differences, the fascicle length change was calculated as the difference between the reference length and the measured fascicle length during the steady-state step.

The equations by Menegaldo et al. [2004] were used to calculate the GM musculotendon complex length change (muscle plus free tendon and aponeurosis in both distal and proximal ends) using ankle and knee joint displacements. The difference between the MTC length change and the fascicle length change, taking the pennation angle into account, was calculated as an estimate of the whole tendon (free tendons plus aponeuroses) length change (Figure 2.2).

The EMG signals were band-pass filtered (20-450 Hz) using the Delsys software, rectified, smoothed (using a 2nd order low pass 5 Hz bidirectional filter) and normalized to their own maximum (EMGmax) during the phase analyzed, using Matlab. To study muscle length changes during periods of muscle activity we determined these from the EMG signal taking into account an electro-mechanical delay estimated at 24ms [de Ruiter et al., 2004; Muraoka et al., 2004]. Foot contact phases were separated by bilateral foot contact and lift-off events. The foot contact phase where the peak EMG amplitude was obtained was used to further analyze the fascicle length data. During this phase, the velocity of fascicle length changes was averaged. Averaged fascicle contraction velocities were compared to zero using a one sample t-test to determine whether shortening or lengthening occurred.

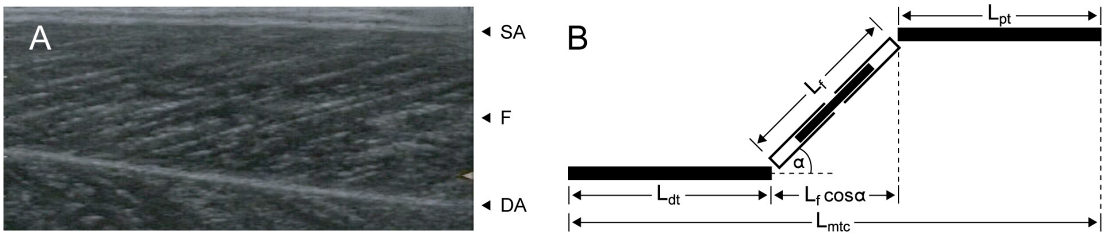

Figure 2.2. (A) Typical sonograph of the gastrocnemius medialis (GM) muscle during the stance phase of stair descent. GM muscle fascicles (F) lie in between and in parallel with thin white lines, which represent inter-fascicular tissue. GM muscle fascicles run between the superficial aponeurosis (SA) and the deep aponeurosis (DA). (B) Schematic representation of the musculotendon model used. L_f is the fascicular length, α is the pennation angle, L_{pt} is the proximal tendon (free tendon and aponeurosis) length, L_{dt} is the distal tendon (free tendon and aponeurosis) length, and L_{mtc} is the musculotendon complex length. L_f and α are measured using ultrasound imaging (see A), L_{mtc} is calculated from the kinematic data from ankle and knee joint (see text). Total tendon length (L_{pt} + L_{dt}) is estimated by the length difference in L_{mtc} and $L_f\cos\alpha$. [Figure adapted from Fukunaga et al. [2001]]

Results

Stair Ascent

Averaged stride cycle time for ascent was 1.3 sec, ranging from 1.03 to 1.42 sec. Results for the ascending trials are shown in Table 2.1 and Figure 2.3. During the stance phase, until the start of the second double support phase, MTC length and fascicle length both increased slightly and in parallel, resulting in a constant tendon stretch of about 1 cm. In the second part of single support stance, the GM muscle became active and the fascicle length remained constant, while MTC length increased slightly, stretching the tendon a little more. After the contra-lateral leg touched down again, the fascicles tended to shorten, while the MTC shortened rapidly, following the burst in GM muscle activity and the decrease in external force on the foot. This fascicle contraction resulted in a peak ankle moment, leading to push-off. During the swing phase, in the absence of external force on the foot, GM muscle fascicle shortening followed the MTC length change, until ankle dorsiflexion stretched the GM muscle fascicles.

The GM muscle activity peaked during the lift-off phase (second double support phase). While the EMG amplitude remained over 30% EMGmax, the GM muscle fascicles shortened by 0.02 cm per 1% of the stride cycle, which was, however, not significantly different from zero (p=0.140). Hence, during lift-off, the GM muscle fascicles can be considered to have contracted almost isometrically.

Table 2.1. Gastrocnemius muscle fascicle length during stair ascent

	Percentage of stride length*				
	0%	14%	50%	63%	100%
Muscle fascicle length (cm)	5.0 ± 0.7	5.1 ± 0.7	5.7 ± 0.8	5.2 ± 0.8	4.9 ± 0.8
Muscle fascicle length change (cm)**	-1.5 ± 0.3	-1.3 ± 0.4	-0.8 ± 0.6	-1.2 ± 0.6	-1.5 ± 0.5
MTC length change (cm)	-0.4 ± 0.3	0.7 ± 0.2	0.6 ± 0.2	-0.8 ± 0.4	-0.4 ± 0.3
Tendon length change (cm)	0.9 ± 0.2	1.2 ± 0.4	1.3 ± 0.5	0.3 ± 0.6	1.0 ± 0.4

Values are means ± SD. MTC, musculotendon complex.

* 0, 14, 50, 63 and 100% correspond, respectively, to touch-down of the analyzed leg, lift-off of the contra-lateral leg, touch-down of the contra-lateral leg, lift-off of the analyzed leg and touch-down of the analyzed leg.

** Difference between measured fascicle length during stair negotiation and measured fascicle length during stance.

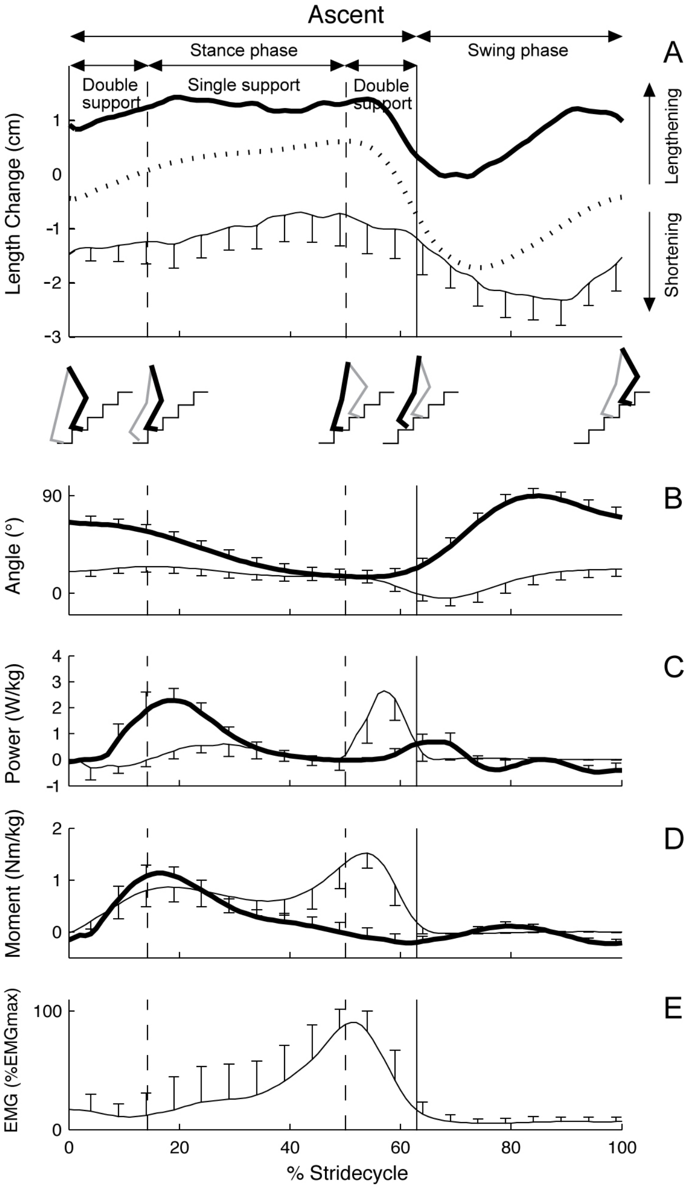

Figure 2.3. (A) Mean changes in GM muscle fascicle length (thin line), musculotendon complex length (dashed line) and tendon length (thick line) during ascent (*n*=10). Standard deviation of the changes in GM muscle fascicle length is shown by error-bars. Values are relative to the reference position (standing still). The percentage of stride length is shown on the x-axis, beginning with touch-down of the left (analyzed) leg. (B) Mean ankle (thin line) and knee (thick line) joint angles during the same stride cycle (*n*=10). (C) Mean ankle (thin line) and knee (thick line) power values during the same stride cycle (*n*=10). (D) Mean ankle (thin line) and knee (thick line) moment values during the same stride cycle (*n*=10). (E) Mean EMG of the GM muscle (*n*=9) during the same stride cycle, normalized to the maximum value (EMGmax) reached during this stride cycle. EMG data from one subject are missing as offline analysis showed that the acquisition system failed during that trial. For clarity error-bars are pointing in one direction only.

Stair Descent

Averaged stride cycle time for descent was 1.2 sec, ranging from 1.15 to 1.45 sec. Results for descending trials are shown in Table 2.2 and Figure 2.4. During the first double support phase, immediately after touch-down, the GM muscle fascicles shortened rapidly, coinciding with a high level of GM muscle activity. At the same time, MTC length increased, resulting in a considerable tendon stretch. After lift-off of the contra-lateral leg, during the single support phase, the muscle fascicles were elongated passively or with very low muscular activity (evidenced by low GM muscle EMG activity), while the tendon shortened. During the second double support stance phase, after touch-down of the contra-lateral leg, GM muscle fascicle length stayed relatively constant, while both the tendon and MTC length decreased rapidly due to knee flexion and ankle plantarflexion, which resulted in lift-off. During the swing-phase the GM muscle fascicles shortened passively as the ankle joint rotated in the plantarflexion direction, following the MTC length, until just before touch-down, when the GM muscle became active and the GM muscle fascicles actively shortened.

The GM muscle was mainly active during the touch-down phase (first double support phase). During this phase, the GM muscle activity remained over 43% EMGmax and the GM fascicles shortened by 0.05 cm per 1% of the stride cycle, which was significantly less than zero ($p<0.005$). Hence during touch-down, the GM muscle fascicles actively shortened even though the MTC lengthened.

Normalization of EMG data to MVC showed that peak EMG value was 64% of MVC during stair ascent and 50% of MVC during stair descent.

Table 2.2. Gastrocnemius muscle fascicle length during stair descent

	Percentage of stride length*				
	0%	13%	50%	63%	100%
Muscle fascicle length (cm)	4.3 ± 0.8	3.5 ± 0.5	5.8 ± 0.8	5.6 ± 0.6	4.3 ± 0.7
Muscle fascicle length change (cm)**	-2.2 ± 0.4	-2.9 ± 0.2	-0.6 ± 0.4	-0.8 ± 0.4	-2.2 ± 0.4
MTC length change (cm)	-1.5 ± 0.2	0.3 ± 0.1	0.8 ± 0.3	-1.1 ± 0.4	-1.4 ± 0.3
Tendon length change (cm)	0.4 ± 0.4	2.6 ± 0.2	1.3 ± 0.4	-0.4 ± 0.4	0.5 ± 0.4

Values are means ± SD. MTC, musculotendon complex.

* 0, 13, 50, 63 and 100% correspond, respectively, to touch-down of the analyzed leg, lift-off of the contra-lateral leg, touch-down of the contra-lateral leg, lift-off of the analyzed leg and touch-down of the analyzed leg.

** Difference between measured fascicle length during stair negotiation and measured fascicle length during stance.

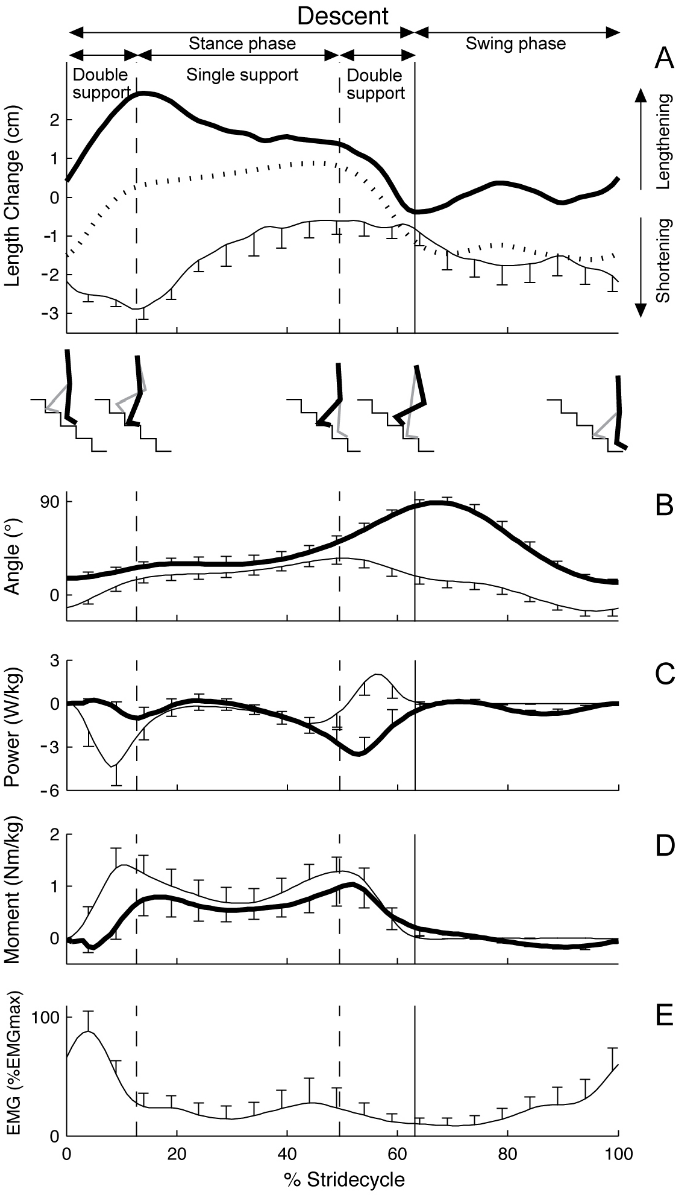

Figure 2.4. (A) Mean changes in GM muscle fascicle length (thin line), musculotendon complex length (dashed line) and tendon length (thick line) during descent. Standard deviation of the changes in GM fascicle length is shown by error-bars. Values are relative to the reference position (standing still). The percentage of stride length is shown on the x-axis, beginning with touch-down of the left (analyzed) leg. (B) Mean ankle (thin line) and knee (thick line) joint angles during the same stride cycle. (C) Mean ankle (thin line) and knee (thick line) power values during the same stride cycle. (D) Mean ankle (thin line) and knee (thick line) moment values during the same stride cycle. (E) Mean EMG of the GM muscle during the same stride cycle, normalized to the maximum value (EMGmax) reached during this stride cycle. For clarity error-bars are pointing in one direction only (n=10).

Discussion

The aim of the present study was to establish the behaviour of human GM muscle fascicles during stair negotiation. The two main findings of the study are the near-isometric behaviour of the GM muscle fascicles during stair ascent and the shortening of the GM muscle fascicles during stair descent. Both findings are in contrast to our hypothesis and show that the length change of muscle fascicles does not follow the MTC length change.

The kinematic results of both the ascending and descending trials show a very small inter-subject variation and are in line with the results from other studies during stair negotiation [Livingston et al., 1991; McFadyen and Winter, 1988; Riener et al., 2002; Yu et al., 1997]. Although the variation was larger for the kinetic results, both the ankle and knee joint moment and power values are also similar to the results from previous studies [McFadyen and Winter, 1988; Riener et al., 2002]. The variation in EMG results is rather large, however, the muscle activity bursts can be distinguished easily, and were, again, similar to the results of other studies [McFadyen and Winter, 1988; Townsend et al., 1978].

Stair Ascent

It has previously been shown that during level walking and walking on an incline (+10%), GM muscle fascicles maintain a near-constant length during the phase in which this muscle is active [Fukunaga et al., 2001; Ishikawa et al., 2005a; Lichtwark and Wilson, 2006]. The present results show, that the length modulation of the MTC (Figure 2.3) is about twice as much as that of the muscle fascicles. Therefore, the MTC length change is not produced by GM muscle fascicle length change alone and substantial tendon length changes occur. The results also show that, when the GM muscle becomes active around the second double support phase during stair ascent, just before lift-off, the fascicles remain near-isometric. This suggests that the GM muscle hardly produces any work. However, Figure 2.3 shows a peak in ankle joint power during this phase. It seems as though the GM muscle fascicles produce force by contracting isometrically and that the tendon recoil is responsible for the peak in ankle joint power, since this caused the rapid MTC shortening. Other ankle plantarflexor muscles, like the soleus muscle, will also contribute to the peak in ankle joint power. This is supported by McFadyen and Winter [1988], who report soleus EMG activity during this part of the stride cycle. Another possibility is that the power at the knee, produced by the knee extensors, is transferred to the ankle through the bi-articular GM muscle. Furthermore, the knee extensors likely produce work during the first double support phase and early single support phase. This notion is supported by the peak in knee joint power seen during the early single support phase in the present study (Figure 2.3) and the peak in vastus lateralis muscle activity around the same time reported in the literature [McFadyen and Winter, 1988; Townsend et al., 1978]. During the landing phase of stair ascent, the foot is placed horizontally on the step and hardly any plantarflexion power is produced until just before lift-off.

Stair Descent

It is generally believed that muscles contract eccentrically and act as brakes to absorb energy when walking down a declined surface. However, Lichtwark and Wilson [2006] showed that, when walking down a slope (-10%), the GM muscle fascicles contract near-isometrically. The present results from stair descent show that although the MTC is stretched and thus absorbs energy as a whole, the GM muscle fascicles actually shorten considerably during touchdown and the first double support phase. This active shortening of GM muscle fascicles will produce energy that will be stored in the tendon, producing an ankle joint moment sufficient to overcome the dorsiflexion acceleration caused by gravity.

In stair descent, heel strike was absent and subjects landed on their forefoot, which makes it a different task than walking down a declined surface. As such, the landing phase of stair descent appears to some extent comparable to counter-movement exercises, such as described by Kawakami et al. [2002]. These authors showed that during the plantarflexion counter-movement, GM muscle fascicles acted almost isometrically, thus allowing storage of elastic energy in the tendon, consequent release of which would enhance exercise performance. However, during counter-movements which are more strenuous, such as drop jumping, the GM muscle fascicles were found to shorten during the touch-down/push-off phase, while the MTC lengthened [Ishikawa et al., 2005b]. This shortening of GM muscle fascicles during MTC stretch was also seen in in-situ experiments in rat muscles [Ettema et al., 1990] and in freely walking cats [Griffiths, 1991] and is similar to GM muscle fascicle behaviour during the touch-down phase of stair descent as seen in the present study. However, this is not always the case. Ishikawa et al. [Ishikawa et al., 2005b] showed that the vastus lateralis muscle fascicles actively lengthened during MTC lengthening in drop jumping, whilst the GM muscle fascicles shortened. In addition, it has been shown that fascicles of the soleus muscle behave in a different way than GM muscle fascicles during locomotion [Ishikawa et al., 2005a]. It can be concluded that muscle fascicular shortening during MTC lengthening can occur during short, intense eccentric actions of muscles with a long compliant tendon. It might be that only bi-articular muscles show this behaviour, but more research is needed to confirm this. This concentric muscle fascicle behaviour during an ‘eccentric’ movement leads to internal energy losses and in this respect is less efficient than isometric behaviour. However, the magnitude of the impact force during stair descent may not allow this efficient isometric strategy. It can, therefore, be hypothesized that the GM muscle is regulating the length of its tendon in order to provide the required stiffness around the ankle in the transition from foot down to single support. Apparently, stiffening of the ankle joint achieved through contracting the muscle is required for a controlled dorsiflexion and this requirement probably dominates over (or cancels out) energetic efficiency criteria in performing the movement.

After touchdown, during single support stance, the GM muscle fascicles are stretched while the EMG shows little or no activity. Some energy that was stored in the tendon is dissipated here, suggested by the decrease in tendon length. Although muscle fascicles can produce considerable forces at low electrical activity when they are lengthened, the decreasing tendon length indicates that the ankle moment is low in this part of the stride cycle and force in the fascicles decreases. The knee extensors are active in this part of the stride cycle [McFadyen and Winter, 1988; Townsend et al., 1978], probably contracting eccentrically and also dissipating energy. During the lift-off phase, the tendon recoils while the fascicles remain at the same length with hardly any muscle activity. This resulted in a shortening of the whole MTC and hence, a small positive power peak.

Methodological considerations in the study include the way that fascicular, tendon and MTC lengths were obtained. The fascicular trajectory was approximated as a straight line, neglecting the slight curvature of the fascicles [Maganaris et al., 1998]. The difference between the two measurement approaches is, however, small (<3%, as estimated in the present study) and falls within the variation of the measurements. Another methodological consideration is that fascicular recordings were taken from one region of the muscle only. However, ultrasound scanning of the gastrocnemius muscle in proximal, central and distal regions during walking and running has shown that fascicle length changes in the central region approximate well the changes occurring throughout the muscle length [Lichtwark and Wilson, 2005]. The calculation of the tendon length change depends on the MTC length change, which was calculated according to Menegaldo et al. [2004] using data of knee and ankle joint angle changes. Other models for prediction of MTC length from joint angles can be found in the literature [Grieve et al., 1978; Hawkins and Hull, 1990; Yuen and Orendurff, 2006]. These alternative models yield similar predictions as the model used.

In conclusion, the present study shows that during stair ascent the GM muscle fascicles contract near-isometrically during the push-off phase, while the MTC shortens providing the ankle joint power needed to elevate the body. During stair descent, the GM muscle fascicles are only active around touch-down and contract concentrically - not eccentrically as expected if the muscle operated as an energy absorbing element. While other muscles, such as the soleus and the knee extensors, may act eccentrically to decelerate the body, the present findings indicate that the GM tendon is stretched and hence stores energy, part of which is dissipated by the elongating muscle fascicles in the mid-stance phase, and by the tendon recoil during lift-off. These findings are in contrast to what would be predicted from joint kinematics only.

Chapter 3

Influence of gait velocity on gastrocnemius muscle fascicle behaviour during stair negotiation

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Abstract

The gastrocnemius medialis (GM) muscle plays an important role in stair negotiation. The aim of the study was to investigate the influence of cadence on GM muscle fascicle behaviour during stair ascent and descent. Ten male subjects (young adults) walked up and down a four-step staircase (with forceplates embedded in the steps) at three velocities (63, 88 and 116 steps/min). GM muscle fascicle length was measured using ultrasonography. In addition, kinematic and kinetic data of the lower legs, and GM electromyography (EMG) were measured. For both ascent and descent, the amount of fascicular shortening, shortening velocity, knee moment, ground reaction force and EMG activity increased monotonically with gait velocity. The ankle moment increased up to 88 st/min where it reached a plateau. The lack of increase in ankle moment coinciding with further shortening of the fascicles can be explained by an increased shortening of the GM musculotendon complex (MTC), as calculated from the knee and ankle angle changes, between 88 and 116 st/min only. For descent, the relative instant of maximum shortening, which occurred during touch down, was delayed at higher gait velocities, even to the extent that this event shifted from the double support to the single support phase.

Introduction

The series elastic component of muscles has been shown to enhance the efficiency of human movement. Examples like jumping, walking and running show that elastic storage and release allow re-use of energy generated by muscle fascicles [Alexander and Bennet-Clark, 1977; Fukashiro et al., 2006; Fukunaga et al., 2001; Ishikawa et al., 2007; Kawakami and Fukunaga, 2006; Kurokawa et al., 2001; Roberts, 2002]. Additionally, the series elastic component allows much faster shortening of the musculotendon complex (MTC) than fascicles alone could realize, resulting in high joint angular velocities [Alexander and Bennet-Clark, 1977; Fukashiro et al., 2006; Kawakami and Fukunaga, 2006; Roberts, 2002]. It seems as though the MTC behaves in the most efficient way for a variety of tasks [Alexander and Bennet-Clark, 1977; Fukashiro et al., 2006; Roberts, 2002].

In Chapter 2, we studied the behaviour of gastrocnemius medialis (GM) muscle fascicles during stair negotiation using ultrasonography. GM muscle fascicles hardly shortened during the push-off phase of stair ascent, while the entire MTC clearly shortened, causing lift-off. This behaviour seems efficient and has also been found in level walking and in walking up or down a slope [Fukunaga et al., 2001; Lichtwark and Wilson, 2006].

In stair descent, the stabilizing function of the GM appeared to take priority over energetic efficiency (Chapter 2). The GM muscle fascicles actively shortened during the touch-down phase of stair descent, while the MTC lengthened. This implies that instead of dissipating energy, muscle fascicles generated energy causing the tendon to stretch (to a greater extent than if the fascicles were to lengthen) and the ankle joint to achieve a high stiffness, probably to provide the required stability. The energy stored in the tendon caused the fascicles to lengthen at a later stage, and the whole MTC to shorten during the tendon recoil at lift-off when the energy was finally dissipated. This process of energy production and subsequent dissipation, in a movement phase requiring only energy dissipation, appears not very efficient.

In Chapter 2, movement was performed at a self-selected pace, and it is known that changes in gait velocity alter the demands placed on the musculoskeletal system. Higher gait velocity is often associated with an increase in step length, resulting in more muscle fascicle shortening and a faster rate of shortening [Gabaldon et al., 2004; Gillis and Biewener, 2001; Hoyt et al., 2005; Roberts et al., 1997]. However, stair walking is a fairly constrained movement in terms of step length as it is imposed by the stairs' dimensions. Nonetheless, with higher stair walking velocities, impact forces and joint moments are expected to be larger, suggesting that the stiffness of the ankle needs to be higher and the muscle fascicles need to shorten more in both ascent and descent. The purpose of the present study was to investigate the influence of gait velocity on the behaviour of GM muscle fascicles during stair negotiation. It was hypothesized that with higher stair walking velocities GM muscle fascicles will shorten more and at a faster rate in both stair ascent and descent.

Methods

Subjects

Ten healthy male volunteers with average age 24.9 ± 3.2 years, height 1.82 ± 0.06 m and mass 79.9 ± 9.1 kg, gave their written informed consent to participate in this study. The same subjects participated in the experiments reported in Chapter 2. The study was approved by the ethics committee of the Institute for Biophysical and Clinical Research into Human Movement at the Manchester Metropolitan University.

Measurements

The experimental set-up has been described in detail previously (see Chapter 2). Participants negotiated a custom-built steel staircase made out of 4 steps. The steps (riser: 17 cm x going: 30 cm x width: 90 cm) were independently mounted on the floor.

A 9-camera VICON 612 system (VICON motion systems Ltd., Oxford UK) was used to acquire kinematic data. Retro-reflective markers were placed on bony landmarks, directly on the skin, or on tight-fitting clothing using double-sided tape. In total, 34 markers were placed on the body according to the standard "plug-in-gait" model of the VICON system implemented in the Bodybuilder software module for 3D segment modelling and calculation of upper and lower limb kinematics. For exact placement of the markers, see Chapter 2.

Forceplates were used to collect kinetic data during stair negotiation. Three force plates (Kistler 9286A, 27 x 52 cm) with built-in amplifiers were embedded in the first three steps (from the ground), and one force plate (Kistler 9253A, 40 by 60 cm) with an external amplifier (Kistler 9865C) was embedded in the floor, in front of the staircase.

The GM muscle fascicle behaviour was assessed in vivo by ultrasound scans recorded in real-time during the stair negotiation trials. For these measurements, an ultrasound system (Aloka SSD-5000, Tokyo, Japan) was used. A linear 7.5 MHz probe (UST-579T-7.5) with 60 mm field of view was tightly secured around the left lower leg in the mid-sagittal plane of the GM muscle with a custom-built fixation device. The fixation device was made of a plastic cast, moulded to fit the general contour of the calf, with a window for the probe. The probe was held rigidly by the cast, which was securely fixed on the calf using Velcro straps. The experimenter supported the probe cable to ensure that no probe-movement, relative to the GM muscle, occurred. Sampling rate was 22 Hz and image resolution was 768 x 576 pixels. The ultrasound scanning was synchronized with the kinematic, kinetic and EMG data using an external trigger.

The electrical activity of the GM muscle of the left leg was recorded using a Bagnoli EMG system (gain 1000, bandwidth 20-450 Hz; Delsys Inc. Boston, MA, USA), with a sampling rate of 2000 Hz. The recording electrodes for the GM muscle were placed proximal to the ultrasound scanning probe in the mid-sagittal plane of the muscle.

Protocol

In order to scale the generic human plug-in-gait model in the VICON software (Oxford Metrics Inc.), anthropometric measurements were taken for each participant before measurements began. Subsequently, the markers, EMG electrodes and ultrasound probe were positioned and data collection was initiated.

Subjects performed three trials of stair descent and three trials of stair ascent, at different gait velocities: 63, 88 and 116 steps per minute, dictated by an audible metronome. Subjects walked barefoot, in a step-over-step fashion. Before the trial started, the subjects stepped on the spot in rhythm with the metronome on top of the platform (stair descent), or on the ground just in front of the ground forceplate (stair ascent), and started every trial with their right foot. The trial ended when the subject was on the top platform, or on the ground, off the forceplate. When a subject wasn't able to perform the trial at the correct pace, a new trial at the same pace was performed.

Data analysis

The phase between the first touchdown point of the left foot and the second touchdown point of the left foot (two steps above/below) was considered a steady-state stride cycle [Andriacchi et al., 1980]. From marker positions and forceplate data, VICON software was used to calculate kinematics and kinetics in 3-D using the plug-in-gait model, while only the sagittal plane information was used for further processing. From the steady-state stride cycle, the kinematic and kinetic data of the ankle and knee were transported from 'VICON workstation' software to Matlab (The Mathworks, Inc, Natick, MA).

The GM muscle fascicle lengths were measured from the recorded ultrasonographic images. On each ultrasound frame from the steady-state stride cycle, GM muscle fascicle length was measured manually using Matlab. Muscle fascicle length was measured using the assumption that the fascicular trajectory was linear. To account for individual differences, the fascicle length change was calculated as the difference between a reference length and the measured fascicle length from the steady-state step. The fascicle length measured on the same occasion, in a standing position, was the reference length for each subject.

The GM musculotendon complex length change (muscle plus free tendon and aponeurosis in both distal and proximal ends) was calculated based on ankle and knee joint rotation, using the equations by Menegaldo et al. [2004].

EMG signals were band-pass filtered (20-450 Hz) by the Delsys system, then rectified, smoothed (2^{nd} order low pass 5 Hz bi-directional filter) and normalized to the maximum EMG measured over all three trials using Matlab. Foot contact phases were separated by bilateral foot contact and lift-off events.

Of interest were the phases where the GM muscle was active, as indicated by the GM EMG, and when the analyzed leg was in the stance phase. Hence, for ascent, GM muscle fascicle shortening was calculated from the peak in the GM EMG trace until lift-off. For descent, GM muscle fascicle shortening was calculated from touch-down until maximum shortening of the fascicles occurred. The root mean square (RMS) from the raw EMG was also calculated for this period.

Statistics

Effect of gait velocity on various variables was tested using Generalised Estimating Equations (GEE) [Liang and Zeger, 1993]. In this regression type of analysis, “gait velocity” was the independent variable and “subject” was the random factor (in order to allow a repeated measures design), with actual gait velocities being different between subjects.

Most data in the results and discussion will be presented as normalized to one full stride cycle, so that variables can be easily compared between gait velocities. However, some data needs to be analyzed in absolute terms; therefore, results and discussion will contain data in normalized and absolute time.

Results

Results focus on differences between the various analyzed trials: 63BPM, 88BPM and 116BPM. For a descriptive analysis of stair negotiation at self-selected gait velocity, the reader is referred to Chapter 2. The actual walking cadences were in good agreement with the target cadences (Table 3.1).

Table 3.1. Target cadences as indicated by a metronome and actual walking cadences as performed by the subjects.

	Metronome speeds		
	63 BPM	88 BPM	116 BPM
Ascent (steps/min)	64.1 ± 2.9	89.0 ± 2.4	115.5 ± 4.5
Descent (steps/min)	65.9 ± 2.7	91.6 ± 3.9	117.7 ± 4.3

Ascent

The GEE analysis showed a significant effect of gait velocity ($p<0.05$) for foot-contact events in normalized time (lift-off for both feet). With higher gait velocities, the relative time (relative to the stride cycle) between touch-down and lift-off became shorter, indicating a relatively shorter stance phase and, hence, a relatively longer swing phase.

Table 3.2. Peak joint moments and ground reaction forces, and values for the amount of GM fascicle shortening and shortening velocity for all three gait velocities, both ascent and descent.

		Metronome speeds		
		63 BPM	88 BPM	116 BPM
Ascent	Peak Knee Moment (Nm/kg) *	1.16 ± 0.13	1.24 ± 0.12	1.36 ± 0.15
	Peak Ankle Moment (Nm/kg) *	1.40 ± 0.25	1.49 ± 0.25	1.51 ± 0.21
	Peak vertical GRF (N/kg) *	1.00 ± 0.06	1.07 ± 0.10	1.12 ± 0.07
	Max. Δ Fascicle shortening (cm) *	0.21 ± 0.35	0.43 ± 0.40	0.62 ± 0.54
	Fascicle shortening velocity (cm/s) *	1.31 ± 1.85	2.34 ± 1.96	4.24 ± 3.63
Descent	Peak Knee Moment (Nm/kg) *	0.53 ± 0.30	0.77 ± 0.25	1.04 ± 0.31
	Peak Ankle Moment (Nm/kg) *	1.19 ± 0.28	1.42 ± 0.31	1.45 ± 0.32
	Peak vertical GRF (N/kg) *	1.15 ± 0.16	1.41 ± 0.24	1.52 ± 0.22
	Max. Fascicle shortening (cm) *	3.05 ± 0.29	3.18 ± 0.31	3.26 ± 0.24
	Fascicle shortening velocity (cm/s) *	5.14 ± 1.91	5.52 ± 2.14	7.93 ± 2.12

* Effect of velocity, $p<0.05$ (GEE)

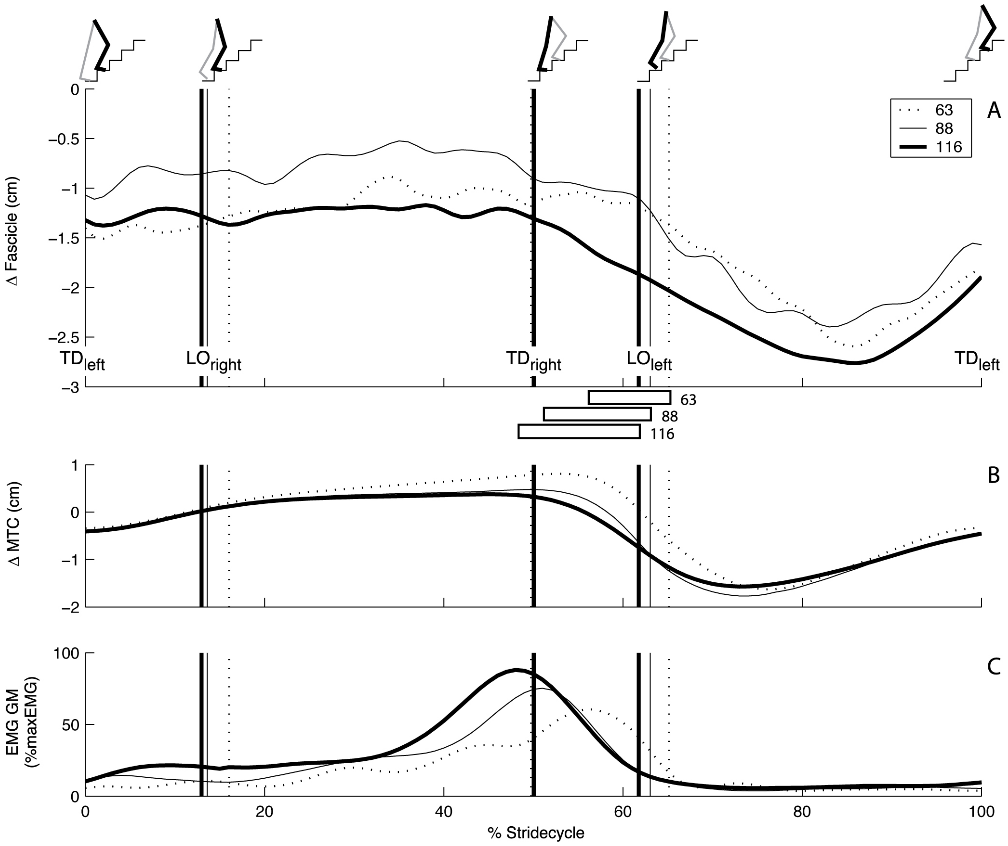

Figure 3.1. Fascicle length change (A), musculotendon complex (MTC) length change (B) and smoothed GM EMG (C) for all three stair ascending velocities: 63 (dotted), 88 (thin) and 116 (thick) steps/min. The fascicular behaviour was analyzed from peak GM EMG until lift-off, this phase was different for all velocities (see bars below the top panel). Vertical lines indicate foot-contact times (lift-off right (LO_{right}), touchdown right (TD_{right}) and lift-off left (LO_{left})) for all three velocities, whereas 0 and 100% of the stride cycle indicate touch-down from the left (analyzed) foot (TD_{left}). EMG data from 1 subject are missing because offline analysis showed that the acquisition system failed during those trials. For clarity, standard deviations are omitted.

The kinetics were also affected by gait velocity (Table 3.2). Peaks in ankle moment and ground reaction force (the second peak during the stride cycle for both variables) occurred during the push-off phase, at the time the GM muscle was active as indicated by the EMG (see Figure 3.1C). The GEE analysis indicated a higher ankle peak moment and ground reaction force peak for higher gait velocities although there were no differences between the two fastest conditions ($p=0.628$ and $p=0.1$, respectively), indicating that a plateau had been reached. Knee moment was also significantly higher with faster gait velocities, and the peak knee moment still increased beyond 88BPM ($p<0.005$).

The effect of gait velocity on GM muscle fascicle behaviour is shown in Figure 3.1A and Figure 3.2A. The GEE analysis revealed an effect of velocity on the amount of fascicular shortening from peak GM EMG until lift-off. The absolute shortening velocity (averaged over the analyzed phase) also increased with gait velocity (Figure 3.2B). With higher gait velocity, there was a greater amount of fascicular shortening and higher

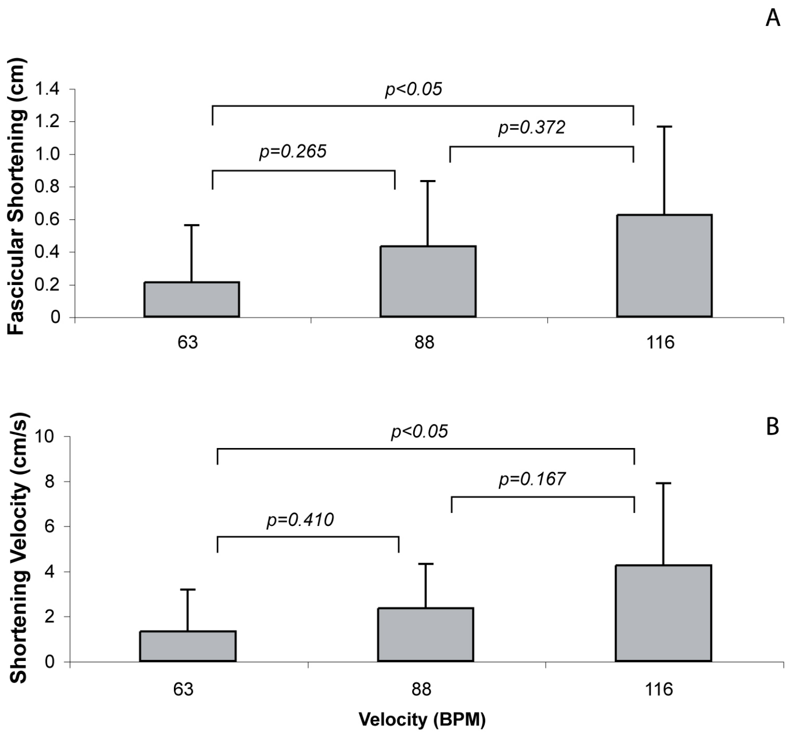

Figure 3.2. GM muscle fascicle shortening (A) and shortening velocity (B) for stair ascent (mean and standard deviations). Analysis was done from peak GM EMG activity until lift-off.

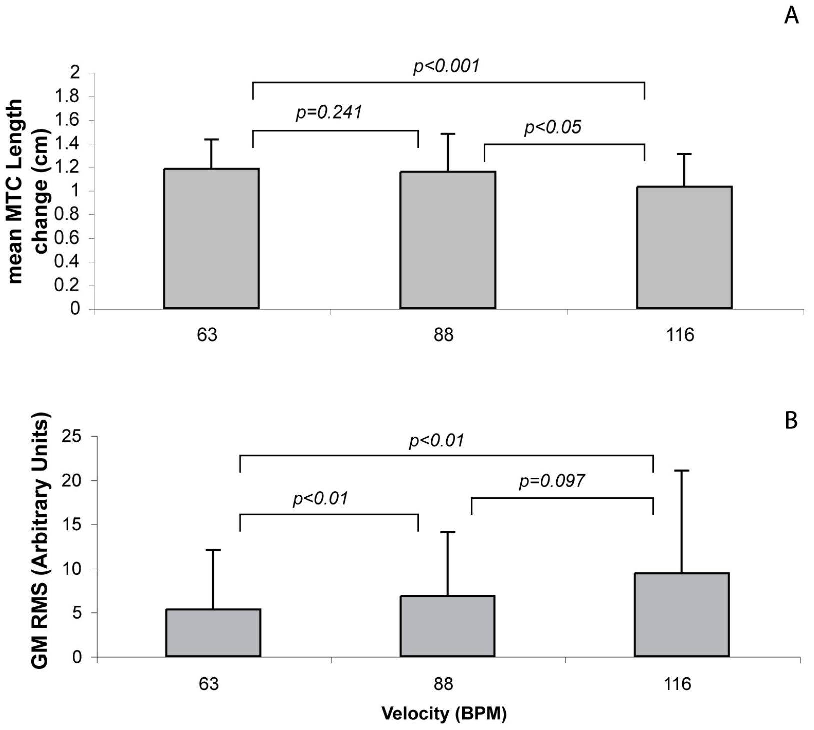

Figure 3.3. Mean musculotendon complex (MTC) length change (A) and root mean square (RMS) values (B) over the phase analyzed (from peak GM EMG until lift-off) in stair ascent (error bars indicate standard deviations).

shortening velocity, although the differences between the 63 and 88BPM conditions and the 88 and 116BPM conditions were not significant on post-hoc testing (Figures 3.2A and B).

The pattern of the MTC length change was little affected by gait velocity (Figure 3.1B), however, the mean MTC length (Figure 3.3A) between EMG peak and lift-off was significantly affected by gait velocity, with a shorter MTC length at higher velocities.

EMG activity increased and peak activity occurred earlier in the stride cycle with higher gait velocities (Figure 3.1C). RMS values increased with gait velocity ($p<0.01$, Figure 3.3B). Peak EMG activity during a trial at a self-selected gait velocity, comparable with the 88BPM trial, was estimated at 64% of a maximal voluntary contraction (MVC), as described in Chapter 2.

Descent

The foot-contact events (lift-off and touch-down for both feet), relative to stride cycle, were similar for all the gait velocities ($p=0.538$). This means that the single and double support phases and the swing phase, were of equal relative length for all the gait velocities.

Kinetics were influenced by gait velocity (Table 3.2) as indicated by GEE. The peaks in ankle and knee moment and in the ground reaction force (at the start of the stance phase at which time the muscle showed substantial EMG activity, see Figure 3.4C) were higher at faster gait velocities. Only the ankle peak moment did not further increase after 88BPM ($p=0.325$).

Shortening of the GM muscle fascicles was influenced by gait velocity (Figures 3.4A and 3.5A, Table 3.2). Maximal shortening, which occurred at the start of the stance phase, increased for higher gait velocities ($p<0.05$). However, there was no statistical difference in maximum shortening between 63BPM and 88BPM ($p=0.362$) or 88BPM and 116BPM ($p=0.635$). The relative timing of this maximal shortening also differed significantly with gait velocity; maximal shortening occurred later in relative time for higher gait velocities ($p<0.005$). The absolute shortening velocity (averaged over the analyzed phase) of the GM muscle fascicles also increased with higher velocities ($p<0.05$, Figure 3.5B). Thus, with higher gait velocities, the GM muscle fascicles shorten more, faster and for a longer period of the stride cycle.

MTC length, at the time of maximum fascicle shortening, was longer at higher gait velocities ($p<0.001$, Figure 3.6A). RMS values of the GM EMG during the analyzed period increased with gait velocity ($p<0.001$, Figure 3.6B), timing characteristics were not affected (Figure 3.4C). Peak EMG activity during a trial at a self-selected gait velocity, comparable with the 88BPM trial, was estimated at 50% MVC, as described in Chapter 2.

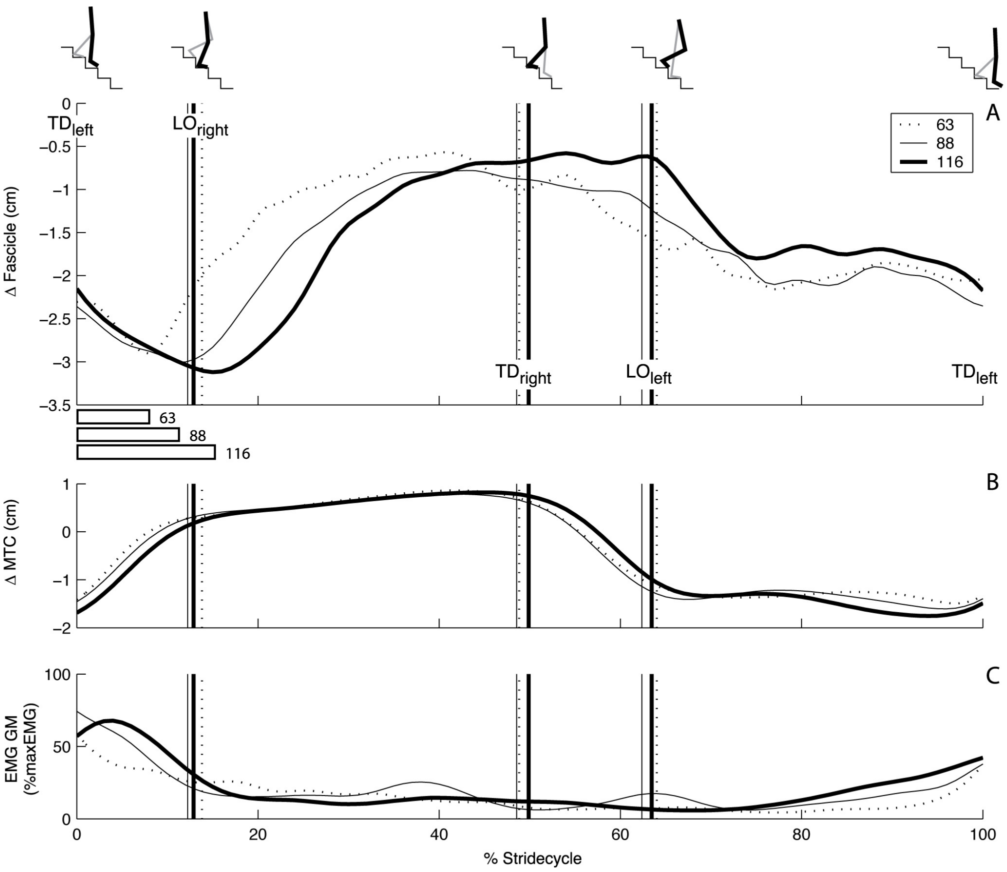

Figure 3.4. Fascicle length change (A), musculotendon complex (MTC) length change (B) and smoothed GM EMG (C) for all three stair descending velocities: 63 (dotted), 88 (thin) and 116 (thick) steps/min. The fascicular behaviour was analyzed from touchdown until maximum GM muscle fascicle shortening, this phase was different for all velocities (see bars below the top panel). Vertical lines indicate foot-contact times (lift-off right (LO_{right}), touchdown right (TD_{right}) and lift-off left (LO_{left})) for all three velocities, whereas 0 and 100% of the stride cycle indicate touch-down from the left (analyzed) foot (TD_{left}). For clarity, standard deviations are omitted.

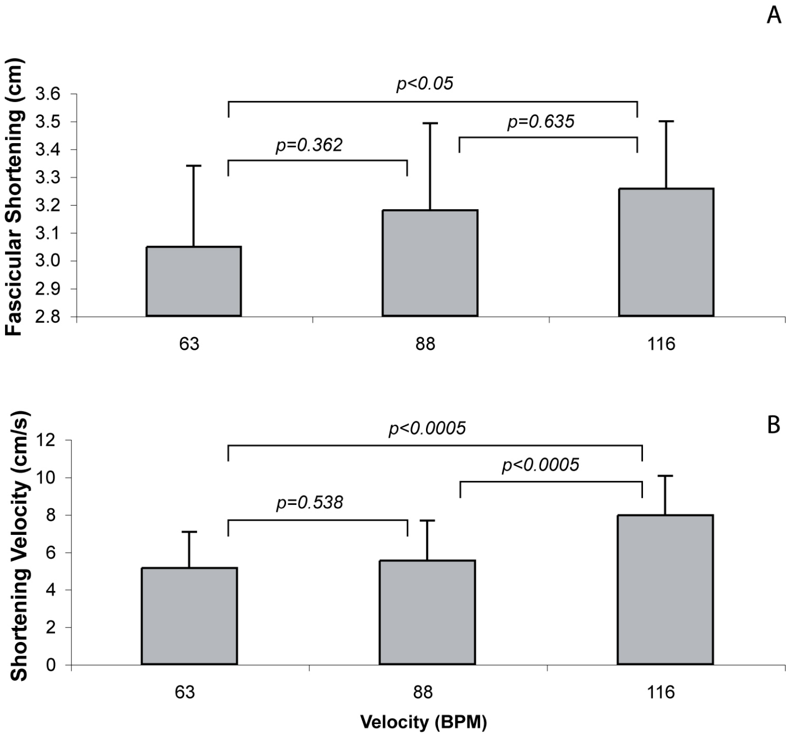

Figure 3.5. GM muscle fascicle shortening (A) and shortening velocity (B) for stair descent (mean and standard deviations). Analysis was done from touch-down until maximum fascicle shortening.

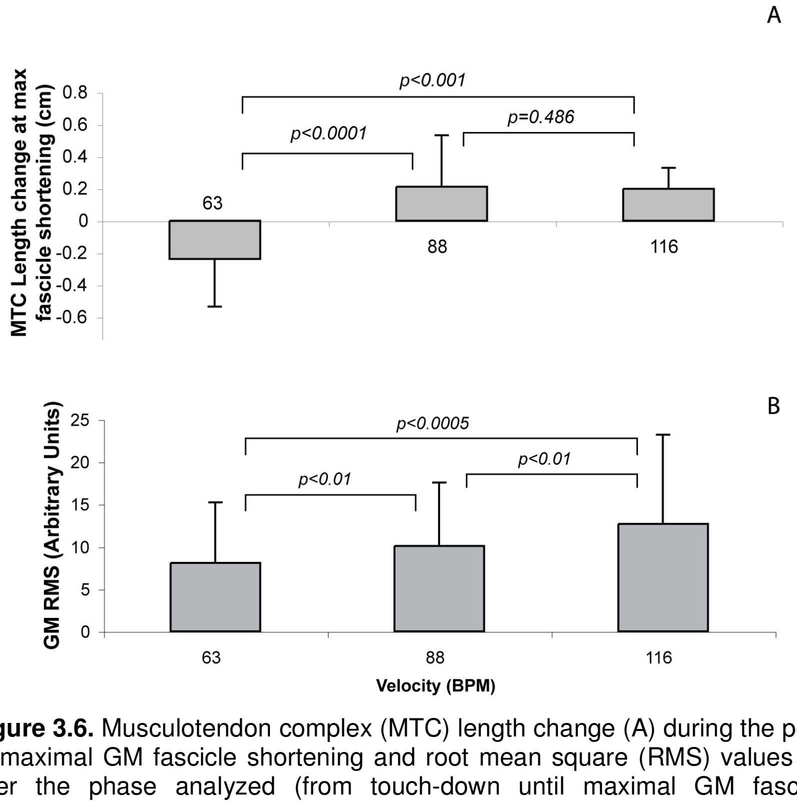

Figure 3.6. Musculotendon complex (MTC) length change (A) during the point of maximal GM fascicle shortening and root mean square (RMS) values (B) over the phase analyzed (from touch-down until maximal GM fascicle shortening) in stair descent (mean and standard deviations).

Discussion

The purpose of the present study was to investigate the influence of gait velocity on the behaviour of GM muscle fascicles during stair negotiation. In line with our hypothesis, the results show an increase in fascicular shortening and rate of fascicle shortening with higher gait velocities in both ascent and descent. Increased fascicular shortening with higher gait velocities is often ascribed to an increase in step length [Gabaldon et al., 2004; Gillis and Biewener, 2001; Hoyt et al., 2005; Roberts et al., 1997]. However, step length is fairly constrained during stair negotiation by the riser and going dimensions of the staircase. Therefore, the general pattern of MTC changes seems similar in all gait velocities. However, in contrast to what we expected, during the phases of interest for both ascent and descent, the MTC length differed between gait velocities. So, increases in fascicle shortening might be ascribed to changes in MTC length or the requirement to produce higher joint moments.

Stair Ascent

The differences between gait velocities in the timing pattern of EMG for stair ascent can be largely explained by the difference in foot-contact times. The higher the gait velocity, the shorter the double support stance phase so EMG activity was shifted earlier in the stride cycle (Figure 3.1C). Peak ankle moment and peak knee moment increased with gait velocity to cause a greater accelerating push off. Peak ankle moment did not increase any further after 88BPM, indicating that a plateau had been reached. Pilot data in our lab (from dynamometer measurements) indicate that peak ankle moment corresponds to ~85% of maximal capacity during stair ascent at 88BPM, while the peak knee moment corresponds to ~50% of maximal capacity. It seems that the peak ankle moment may be closer to the maximal capacity, whilst the peak knee moment is well within maximal capacity and this may explain why peak knee moment increased beyond 88BPM, while ankle moment did not.

Gait velocity had the expected effect on GM muscle fascicle shortening during ascent. At higher gait velocities, more shortening took place from peak GM EMG until foot lift-off. The velocity of fascicle shortening (the slope of the fascicle length data) was also higher at higher gait velocities; according to the force-velocity relationship, it is more difficult to produce a higher force at this higher shortening velocity. Both increased shortening and increased shortening velocity may have required the higher activation that was reflected in higher EMG RMS values (Figure 3.3B) [Carpentier et al., 1996; Hill, 1953]. The increase in shortening with gait velocity agrees well with data of Chino et al. [2008], who showed more GM muscle fascicle shortening and higher shortening velocity at higher angular velocities, measured on a dynamometer using ultrasonography.

However, during contractions on a dynamometer, the MTC length is controlled, while during stair ascent, the MTC length was shorter at higher gait velocities during the phase of fascicle shortening (Figures 3.1B and 3.3A). A higher plantarflexion ankle angle caused the MTC length to be shorter.

Stair Descent

The general pattern of the MTC length change seems quite similar between gait velocities (in relative time, Figure 3.4B). However, during the phase of interest, small changes in MTC length occurred between gait velocities, mainly caused by the ankle angle which was more plantar flexed at higher velocities. This indicates that depending on velocity, different strategies are employed to negotiate a given mechanical environment, just as in stair ascent. Nevertheless, the relative foot-contact times did not change with velocity, just as the general pattern of the GM EMG. The pattern of the ankle and knee moments and the ground reaction forces were also similar between gait velocities. At higher gait velocities, the peak ankle moment and peak knee moment were higher. Apparently, the joints were stiffened to prevent too much flexion at higher gait velocities. However, the ankle moment did not seem to increase further between 88 and 116BPM, indicating that a plateau had been reached. Instead, the knee moment increased further to control the descent. Reeves et al. [2008] recently showed that peak moment during stair descent at a self-selected speed (similar to 88BPM) can be as high as 74% of maximal capacity at the ankle and ~30% of maximal capacity at the knee. Therefore, it seems that there is much more room for the knee moment to increase further when needed.

Gait velocity did have an effect on GM muscle fascicle behaviour, with more shortening at higher gait velocities. More shortening will lead to a larger tendon stretch, and this will produce a higher force in the tendon, resulting in a higher peak moment for the ankle. When going down stairs at a faster pace, peak ground reaction forces are higher caused by the higher downward velocity of the body centre of mass which needs to be counteracted at every step. Therefore, joint moments need to be higher, which is achieved, at least in part, by the increased shortening of the GM muscle fascicles.

Besides shortening more, the fascicles also shortened faster at higher gait velocities ($p<0.001$, Table 3.2). According to the force-velocity relationship [Hill, 1953], fascicles can produce less force at higher shortening velocities. This would mean that, at higher stair descent velocities, the GM muscle fascicles are in a less optimal state to produce force, while the ankle joint moment suggests that they are probably producing the same, or more force than at lower gait velocities. The latter statement is in agreement with the theoretical point of view that when the fascicles are shorter, the tendon is stretched more (when the MTC length is the same), and thus, the fascicles must produce more force. For producing the same, or more force at higher shortening velocities, muscle activation should increase, which is the case for the GM EMG (Figure 3.4C), as indicated by the RMS values that are higher at higher gait velocities (Figure 3.6B). This effect, which was significant over, and between, all gait velocities, seems to explain the increase in GM muscle fascicle shortening. The increased shortening at higher gait velocities is in agreement with dynamometer studies [Chino et al., 2008]. However, the coincident shortening of fascicles and lengthening of the MTC are not likely to occur on a dynamometer.

An interesting result is the gait velocity-dependent shift in relative timing of maximum GM fascicle shortening. With higher descent velocities, the point at which maximum shortening took place, occurred later in the stride cycle. At 63BPM, this event took place well within the first double support stance phase, just after touch down, while it took place in the single support stance phase for the 116BPM, where the ankle and knee kinematics were quite different. There was no difference in relative timing of the peak ankle moment (p=0.804). So, this shift of maximal shortening is not clearly linked to either a kinematic or a kinetic event. To further investigate this, the results were scaled back to absolute times. Remarkably, in absolute time, the timing of maximum GM muscle fascicle shortening was the same at all gait velocities (Figure 3.7), even though there was a difference in kinematics and kinetics. We expected that maximal fascicle shortening would occur earlier in the faster gait (and fascicle shortening) velocities, but it did not. A likely reason for this surprising result may be that at higher shortening velocities the fascicles also shortened more, requiring some extra time to reach their maximum shortening.

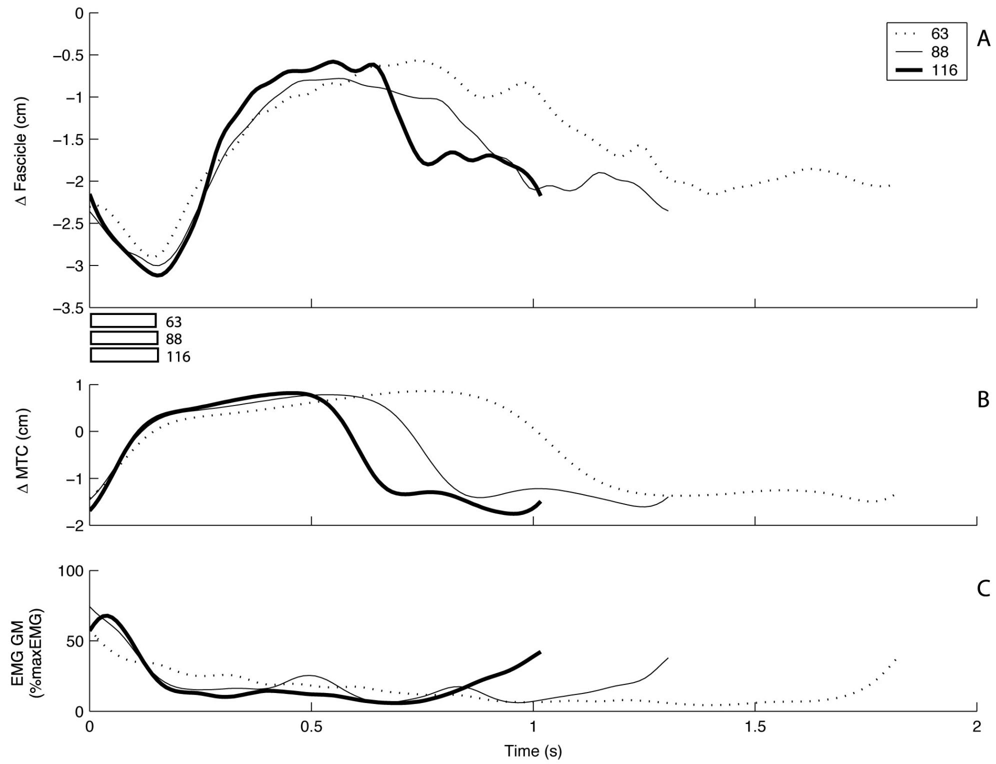

Figure 3.7. Fascicle length change (A), musculotendon complex (MTC) length change (B) and smoothed GM EMG (C) in absolute time for all three stair descending velocities: 63 (dotted), 88 (thin) and 116 (thick) steps/min. The fascicular behaviour was analyzed from touchdown until maximum GM muscle fascicle shortening, this phase was the same for all velocities (see bars below the top panel). The start and endpoints of the individual lines indicate touchdown from the left (analyzed) foot. For clarity, standard deviations are omitted.

The discussion of the results presented above, for ascent and descent, shows, that the increase in GM fascicular shortening with gait velocity can be explained by higher GM activation and that this causes the ankle moment to increase with gait velocity. The fascicle shortening does not show a difference between 63 and 88BPM or 88 and 116BPM on post-hoc analysis (Figures 3.2A and 3.5A), probably because of the large variation, but results suggest that the effects are monotonic over the range of gait velocities tested, just as the effects seen in GM RMS values (Figures 3.3B and 3.6B). However, for ascent and descent, the ankle moment does not increase beyond 88BPM (p=0.628 and p=0.211 respectively), showing a “ceiling effect”. As mentioned above, the ankle moment has reached a value close to maximum, while the knee moment can still increase. But why is this not shown in the fascicle shortening? A possible explanation for this may be found in the MTC length. During the phase of GM muscle fascicle shortening (for ascent), which differed between gait velocities, the MTC length was shorter at higher gait velocities. This effect was not seen between 63 and 88BPM but it was seen between

88 and 116BPM, in apparent contradiction with the above “ceiling effect” (Figure 3.3A). So, the fascicle shortening increased from 88 to 116BPM, where the ankle joint moment did not. This can be explained by the fact that the MTC was shorter in the 116BPM condition; therefore, the fascicles would have to shorten more to achieve a similar muscle force. The MTC length at maximal fascicle shortening during the touch-down phase of descent was longer at higher gait velocities. This was caused by the shift in timing of maximal fascicle shortening (maximum fascicle shortening occurred later in the stride cycle with increasing gait velocity), since the MTC length is increasing in this part of the stride cycle. The mechanism accounting for the “ceiling effect” (EMG and ankle joint moment reached a plateau while the fascicles seemed to shorten further) observed during stair descent remains unknown, it is likely that other muscles play a role here.

The “ceiling effect” and the shift in timing of maximal fascicle shortening indicate that the relationships between fascicle shortening and kinetics or kinematics are not straightforward. This is further compounded by the pennation angle and the non-linear properties of tendon. Thus the behaviour of the muscle fascicles is hard to predict, even with information about muscle activation (EMG), kinetics and kinematics not in the least because other muscles, like the Soleus, can have a plantar-flexing effect on the ankle joint. This will probably affect the relationship between GM muscle fascicle behaviour and ankle moment, especially given that the ratio of activation between the Soleus muscle and the GM muscle is likely to change with movement velocity, with the GM muscle activated more at higher movement velocities [Tamaki et al., 1997]. Furthermore, other thigh and leg muscles are also greatly involved in stair negotiation [Larsen et al., 2007; McFadyen and Winter, 1988] and their actions and relative contributions may also change with velocity.

However, GM muscle function may be crucial in stair negotiation. Where the ankle moment can rise to ~75% MVC during stair negotiation [Reeves et al., 2008], it is the GM muscle that is expected to generate any modulation of this moment [Maganaris et al., 2006]. This is supported by the more pronounced EMG modulation of the GM muscle compared to the soleus muscle in both stair ascent and descent [McFadyen and Winter, 1988].

When we compare the results from our previous study into GM muscle fascicle behaviour during stair negotiation at a self-selected speed (Chapter 2), then, the amount of GM muscle fascicle shortening for ascent agrees well with the shortening found in the present study, slightly more shortening than 88BPM but less shortening than 116BPM. Note, however, that the shortening velocity at a self-selected speed was not significantly different from zero (see Chapter 2). Also in the present study, shortening velocity was not in all conditions significantly different from zero. The monotonic and significant change with ascent velocity however suggests that some fascicle shortening is occurring in this push-off phase. For descent, the results are slightly different. The GM muscle fascicle shortenings at 63, 88 and 116BPM measured in the present experiment, were all larger than the amount of shortening at a self-selected gait velocity. This indicates a higher efficiency of the self-selected gait velocity, which was also associated with slight differences in knee and ankle kinematics. It seems as though the shortening of the GM muscle fascicles and thus the stiffness of the MTC both increase with gait velocity at ascent and descent. This suggests that the efficiency of the GM muscle function decreases when gait velocity is increased.

Conclusion

GM muscle fascicle shortening and shortening velocity both increased for higher stair gait velocities during ascent and descent. More fascicle shortening caused the ankle joint moment to increase in order to produce a higher ankle plantarflexion moment at stair ascent lift-off and in order to control the dorsiflexion movement at stair descent touch-down. The results indicate that the relationships between fascicle shortening and kinetics or kinematics are not straightforward due to the fact that forces are transferred by a non-linear stiffness.

Chapter 4

Influence of step-height and body mass on gastrocnemius muscle fascicle behaviour during stair ascent

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Abstract

To better understand the role of the ankle plantarflexor muscles in stair negotiation, we examined the effects of manipulation of kinematic and kinetic constraints on the behaviour of the gastrocnemius medialis (GM) muscle during stair ascent. Ten subjects ascended a four-step staircase at four different step-heights (changing the kinematic constraints): standard (17cm), 50% decreased, 50% increased and 75% increased. At the standard height, subjects also ascended the stairs wearing a weighted jacket, adding 20% of their body mass, (changing the kinetic constraints). During stair ascent, kinematics and kinetics of the lower legs were determined using motion capture and ground reaction force measurements. The GM muscle fascicle length was measured during the task with ultrasonography. The amount of GM muscle fascicle shortening increased with step-height, coinciding with an increase in ankle joint moment. The increase in body mass resulted in an increased ankle joint moment, but the amount of GM muscle fascicle shortening during the lift-off phase did not increase, instead, the fascicles were shorter over the whole stride cycle. Increasing demands of stair ascent, by increasing step-height or body mass, requires higher joint moments. The increased ankle joint moment with increasing demands is, at least in part, produced by the increase in GM muscle fascicle shortening.

Introduction

Stair ascent is a common, yet strenuous, task. Besides forward propulsion, extra energy production is needed to elevate the body upwards. The muscles crossing the knee and ankle joint are responsible for most of the work done during stair ascent [McFadyen and Winter, 1988; Spanjaard et al., 2007a]. Joint power reaches about the same absolute level at both joints [Riener et al., 2002; Spanjaard et al., 2007a], however, the ankle plantarflexor joint moment is much higher relative to its maximum (~85%) than the knee extensor joint moment (~50%) [Reeves et al., 2007]. This indicates that the ankle joint plays a crucial and possibly limiting role in stair ascent.

The triceps surae complex, comprising the two gastrocnemius muscle heads and the soleus muscle, is the major muscle group responsible for generating ankle plantarflexion moment. For ankle plantarflexor moments higher than 70% of MVC, the soleus muscle is maximally activated [Maganaris et al., 2006], and thus only the gastrocnemius muscle contributes to the production of higher plantarflexor moments by increasing its activation level and contractile force. Hence, gastrocnemius muscle function may be crucial in situations where high ankle plantarflexion moments relative to maximal capabilities are required, such as during stair ascent as shown by Reeves et al. (2007). This notion is also supported by the finding of a marked modulation of the electromyographic (EMG) amplitude of the gastrocnemius medialis (GM) muscle, compared with a much less pronounced modulation of the EMG amplitude of the soleus muscle during stair ascent in an earlier study [McFadyen and Winter, 1988], highlighting the importance of studying the contractile behaviour of the gastrocnemius muscle in this task.

To gain insight into the way the contractile element in the gastrocnemius muscle operates in stair ascent, the behaviour of the muscle fascicles can be obtained using ultrasonography, and related to joint moment and joint kinematic measurements. An earlier study (Chapter 3) using these techniques indicated that an increase in the demands of stair ascent, by means of increasing gait velocity, required the GM muscle fascicles to shorten more during the lift-off phase. This was associated with an increase in ankle joint moment up to a gait velocity of 88 steps/min. Increasing gait velocity further caused additional GM muscle fascicle shortening, but no further increase in ankle joint moment. In addition, the GM musculotendon complex (MTC) shortened more with increasing gait velocity, due to changes in joint angle patterns, despite the invariant dimensions of the steps in all trials.

To better understand the complex relation between muscle fascicle behaviour and joint biomechanics in stair ascent, in the present study we altered the demands of stair ascent in two different, physiologically relevant, ways: by manipulating body mass and step-height.

Increasing body mass is relevant to real life situations in which a greater total body mass relative to the proportion of lean body mass is present, for example in obese people, older adults and pregnant women. In these conditions, the extra ground reaction forces applied to elevate the body require an increase in the overall support moment, but this moment could be distributed differently between the knee and ankle joints. In addition, differences in joint kinematics might occur despite constant step dimensions, as shown when manipulating gait velocity (Chapter 3). These factors make it difficult to predict the behaviour of the MTC and muscle fascicles with changes in body mass.

On the other hand, experimental manipulation of step-height simulates the real-life situation of negotiating staircases in different locations, for example, stairs with higher steps are in private dwellings and with lower steps in public places [Roys, 2001]. In addition to changes in joint kinematics, higher steps require greater leg joint moments [Livingston et al., 1991; Riener et al., 2002], which complicates the prediction of muscle fascicle behaviour from MTC length changes with alteration in step-height.

Therefore, the aim of the present study was to examine the effects of manipulating body mass and step-height in relation to joint kinetics and kinematics during stair ascent on the GM muscle behaviour.

Methods

Subjects

Ten healthy male volunteers (age: 24.9 ± 3.2 years, height: 1.82 ± 0.06 m and mass: 79.9 ± 9.1 kg), gave their written informed consent to participate in this study. The same subjects participated in previously reported studies (Chapters 2 and 3). The ethics committee of the Institute for Biomedical Research into Human Movement & Health at the Manchester Metropolitan University approved the experimental procedures.

Measurements

The experimental set-up has been described in detail previously (Chapter 2). Subjects ascended a custom-built steel staircase of four steps, barefooted, in a step-over-step manner. The steps were independently mounted on the floor and their height could be altered by using purpose-built metal frames. The tread and width of the steps were kept constant: 280 mm x 900 mm respectively. The riser was set at four heights: standard height (170 mm), 50% decreased height (85 mm), 50% increased height (255 mm) and 75% increased height (297.5 mm).

The subjects were tested on four separate days, one testing session per step-height. When they felt comfortable with the ascent of the staircase (at that specific height), subjects performed one stair ascent at a predefined gait velocity, dictated by a metronome, set at 88 beats per minute (BPM) (which has previously been shown to be close to the self-selected gait velocity in these subjects (Chapter 3)). The measurement was repeated if the cadence of the subject did not correspond to the beats of the metronome (as observed by the experimenter). The trial in which the ascending cadence corresponded best with the metronome was used for analysis. Furthermore, at the standard step-height the subjects performed extra trials. Body mass was increased with 20% by wearing a custom-made jacket filled with pieces of lead distributed uniformly over the trunk. The jacket was secured tightly around the shoulders and waist of the subject without interfering with the rest of the equipment. With the added mass, subjects performed an ascent at 88BPM. In all trials, subjects stood still before the ground forceplate and started the ascent after placing their right foot on the ground forceplate.

Kinetic data during stair ascent were collected using four force plates. Three force plates (Kistler Z17068, 270 x 500 mm) were embedded in the first three steps (from the ground), and one force plate (Kistler 9253A, 400 by 600 mm) was embedded in the floor, in front of the staircase. A 9-camera VICON 612 system (VICON motion systems Ltd., Oxford UK) was used to capture kinematic data. Retro-reflective markers were placed over bony landmarks using double-sided tape. In total, 34 markers were placed on the body according to the standard “plug-in-gait” model of the VICON system. The exact placement of the markers is described elsewhere (Chapter 2).

GM muscle fascicle behaviour was obtained from ultrasound scans recorded at 22 Hz during the trials. For these measurements, a linear 7.5 MHz probe (UST-579T-7.5, Aloka SSD-5000, Tokyo, Japan) with 60 mm field of view was tightly secured around the left lower leg in the mid-sagittal plane of the GM muscle with a custom-built fixation device. The ultrasound scanning was synchronized with the VICON system using an external trigger.

A Bagnoli EMG system (Delsys Inc. Boston, MA, USA) was used to record the electrical activity of the GM muscle of the left leg. The recording electrodes were placed proximal to the ultrasound scanning probe in the mid-sagittal plane of the muscle. The sampling frequency of the EMG recordings was 2000 Hz.

Data analysis

From every trial, the stride cycle between the first touchdown of the left foot (on the first step) and the second touchdown of the left foot (two steps above) was analyzed. VICON software was used to calculate kinematics and kinetics in 3-D, from marker positions and forceplate data, using the “plug-in-gait” model. Only the sagittal plane information was further analyzed.

GM muscle fascicle length was measured manually using Matlab (The Mathworks, Inc, Natick, MA), assuming that the fascicular trajectory was linear. To account for individual differences, the fascicle length change was calculated and is presented as the difference between the length measured in a standing position and the measured fascicle length during the analyzed stride cycle.

GM MTC length change (muscle plus free tendon and aponeurosis in both distal and proximal ends) was estimated using the equation by Menegaldo et al. [2004], taking the ankle and knee joint angles as input. EMG signals were band-pass filtered (20-450 Hz) by the Delsys system, then rectified and smoothed (2^{nd} order low pass 5 Hz bi-directional filter) using Matlab.

The main phase of interest was the period during stance phase where the GM muscle was active. This phase started with the peak of the smoothed EMG and ended with lift-off. For this phase, the amount of GM muscle fascicular shortening, fascicle shortening velocity, MTC length change and the root mean square (RMS) of the raw EMG were calculated. Repeated measures ANOVA and Student t-tests were used to statistically analyze the influence of step-height and added body mass, respectively. For step-height, post-hoc paired t-tests with Bonferonni correction were used to compare adjacent levels.

Results

Step-height

Step-height had no effect on foot contact-times (p=0.862), indicating that the relative duration of stance and swing phase did not change with step-height. However, in spite of the metronome, participants walked at a slightly lower cadence when step-height increased (Table 4.1).

The kinetics changed with step-height; ankle and knee joint moments are shown in Figure 4.1, peak values are given in Table 4.1. During the phase in which the GM muscle was active (according to EMG data) the ankle moment increased with increasing step-height. The knee joint moment (the initial peak in mid-stance) also increased with step-height and presented a secondary peak (Figure 4.1B). The outcome of post-hoc tests is presented in Table 4.1.

We found a main effect of step-height on the amount of GM muscle fascicle shortening from peak EMG to lift-off, with more shortening for increased step-height. (Figure 4.2A and Table 4.1). The shortening velocity of the GM muscle fascicles also increased with increasing step-height.

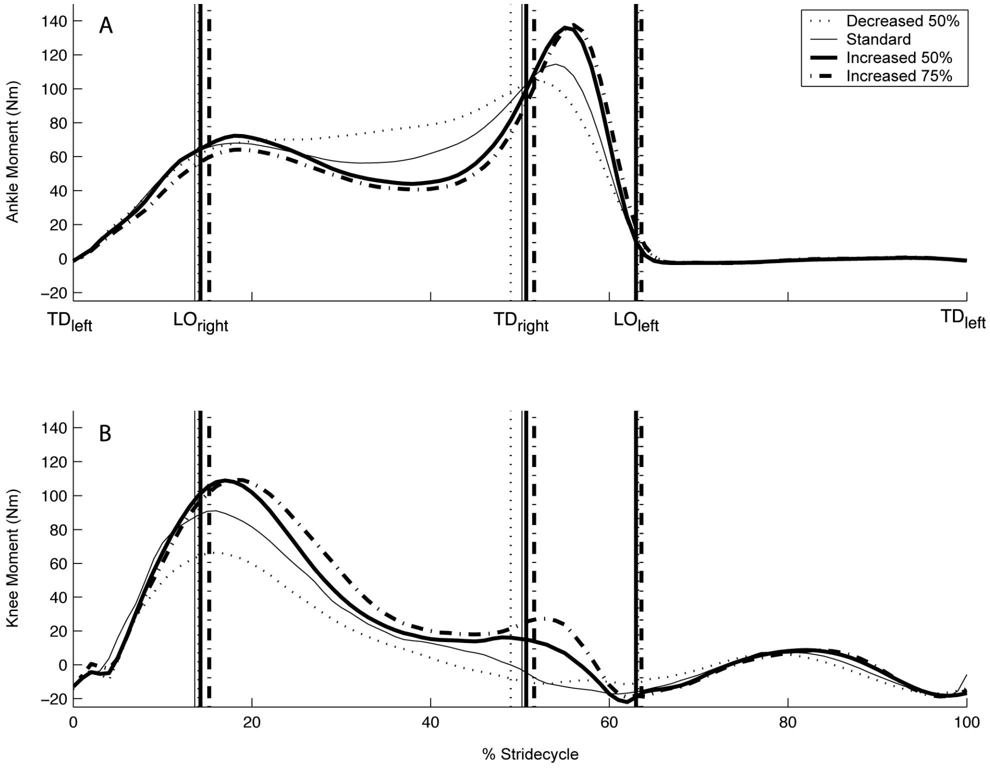

Figure 4.1. Ankle joint moment (A) and knee joint moment (B) of one full ascending stride cycle for all four step-heights: Decreased 50% (dotted), Standard (thin), Increased 50% (thick) and Increased 75% (thick and dashed). Vertical lines indicate foot-contact times (lift-off right (LO$_{right}$), touchdown right (TD$_{right}$) and lift-off left (LO$_{left}$)) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD$_{left}$). For clarity, standard deviations are omitted.

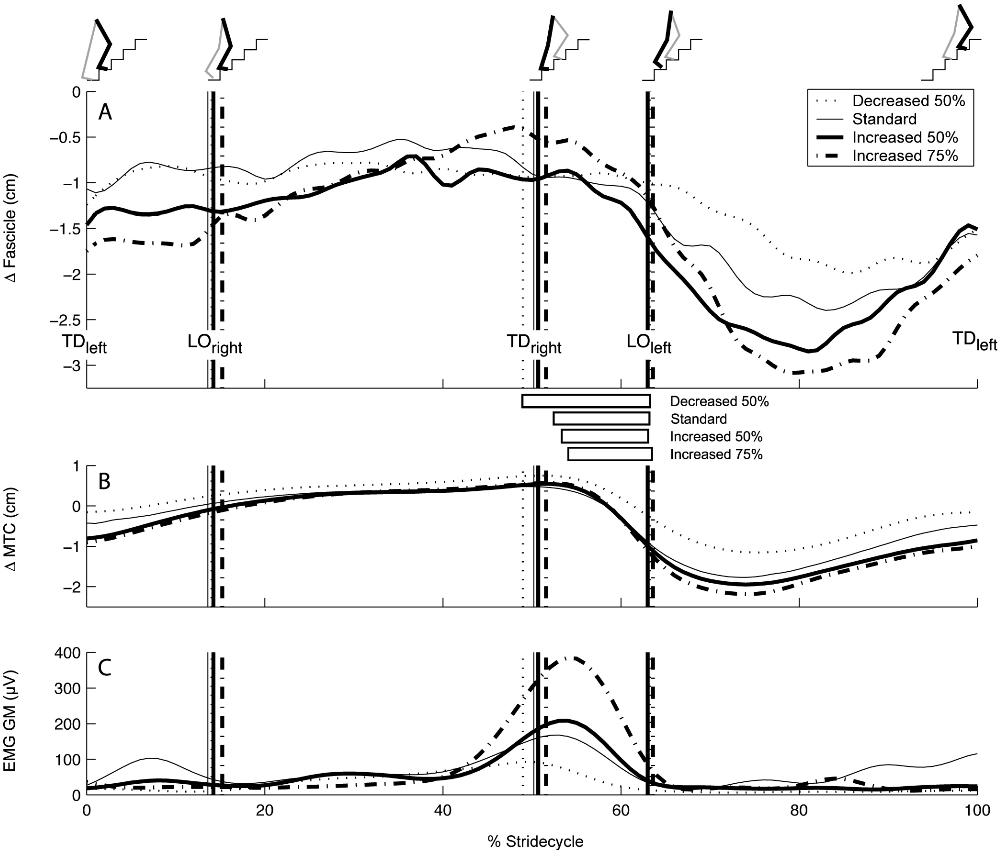

Figure 4.2. Fascicle length change (A), musculotendon complex (MTC) length change (B) and smoothed GM EMG (C) of one full ascending stride cycle for all four step-heights: Decreased 50% (dotted), Standard (thin), Increased 50% (thick) and Increased 75% (thick and dashed). The fascicular behaviour was analyzed from peak GM EMG until lift-off (horizontal bars indicate the analyzed phase per conditions). Vertical lines indicate foot-contact times (lift-off right (LO_{right}), touchdown right (TD_{right}) and lift-off left (LO_{left})) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD_{left}). For clarity, standard deviations are omitted.

The kinematics were greatly influenced by step-height, as reflected in the MTC data (Figure 4.2B). With increasing step-height, the MTC was shorter during the analyzed phase, mainly due to an increased knee flexion angle. The ankle angle showed a much greater range of motion with increasing step-height; more dorsiflexion in the stance phase and more plantarflexion in the swing phase. During the phase analyzed (from peak GM EMG (Figure 4.2C) to lift-off), the MTC was significantly shorter at larger step-heights (Table 4.1).

RMS values of the GM EMG data (Table 4.1) showed that muscle activation increased with increasing step-height.

Table 4.1. Measured and calculated parameters during stair ascent at different step-heights. Values are means ± standard deviations (N=10).

	Step-height					
	Decreased 50%	Standard	Increased 50%	Increased 75%	p-value	Post-hoc ($p < 0.0167$)
Cadence (st/min)	89 ± 5	89 ± 2	86 ± 4	83 ± 3	0.004	
knee moment peak (Nm)	71.1 ± 17.2	98.9 ± 13.1	113.6 ± 21.7	117.0 ± 16.5	0.001	-50 < standard
2^{nd} ankle moment peak (Nm)	106.2 ± 21.0	120.5 ± 33.3	138.7 ± 25.9	142.4 ± 16.3	0.001	standard < +50
GM fascicle shortening (cm)	0.02 ± 0.36	0.46 ± 0.40	0.49 ± 0.64	0.59 ± 0.47	0.026	
GM fascicle shortening velocity (cm/s)	0.2 ± 1.6	2.5 ± 1.9	3.3 ± 4.6	4.4 ± 3.0	0.010	-50 < standard
MTC length change (cm)	0.3 ± 0.2	-0.2 ± 0.2	-0.1 ± 0.2	-0.2 ± 0.2	0.001	
GM EMG RMS (µV)	46 ± 34	101 ± 90	110 ± 59	259 ± 174	0.001	

Added Mass

With 20% extra body mass, participants ascended stairs with a relatively longer double support stance phase ($p<0.005$), while step frequency was not affected (Table 4.2). The ankle and knee joint moments increased with added mass (Figure 4.3 and Table 4.2).

GM muscle fascicle shortening and shortening velocity were both not influenced by the added body mass (Figure 4.4A and Table 4.2). The MTC length was similar between the standard and added mass conditions (Figure 4.4B and Table 4.2) during the phase analyzed (Figure 4.4C). No change in RMS EMG values was found with added body mass.

Table 4.2. Measured and calculated parameters during stair ascent for 2 mass conditions. Values are means ± standard deviations (N=10).

	Body mass		
	standard	added 20%	p-value
Cadence (st/min)	89 ± 2	88.2 ± 2	0.469
knee moment peak (Nm)	98.9 ± 13.1	116.6 ± 24.3	0.008
2^{nd} ankle moment peak (Nm)	120.5 ± 33.3	137.4 ± 37.2	0.001
GM fascicle shortening (cm)	0.46 ± 0.40	0.40 ± 0.49	0.747
GM fascicle shortening velocity (cm/s)	2.5 ± 1.9	1.6 ± 2.1	0.345
MTC length change (cm)	-0.2 ± 0.2	-0.1 ± 0.3	0.240
GM EMG RMS (µV)	101 ± 90	112 ± 82	0.646

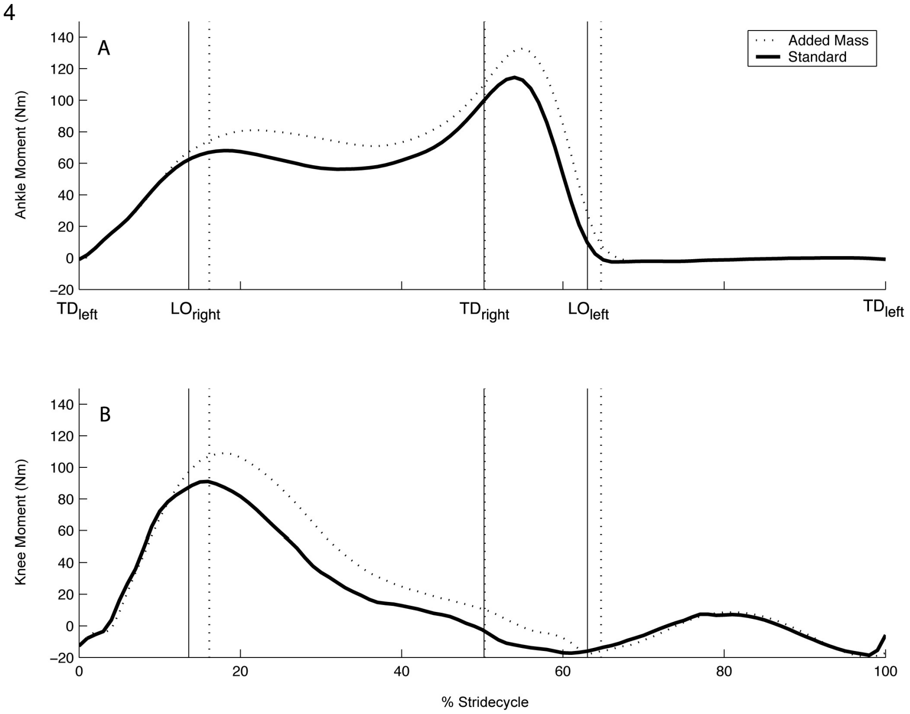

Figure 4.3. Ankle joint moment (A) and knee joint moment (B) of one full ascending stride cycle for standard stair ascent (solid) and stair ascent with 20% extra body mass (dotted). Vertical lines indicate foot-contact times (lift-off right (LO$_{right}$), touchdown right (TD$_{right}$) and lift-off left (LO$_{left}$)) both conditions, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD$_{left}$). For clarity, standard deviations are omitted.

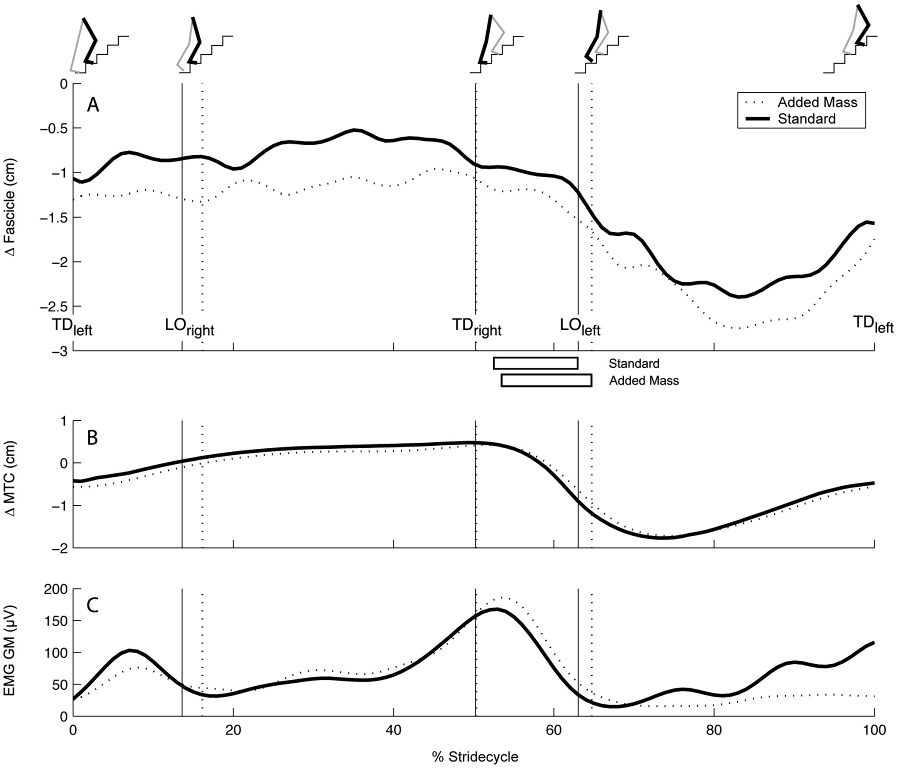

Figure 4.4. Fascicle length change (A), musculotendon complex (MTC) length change (B) and smoothed GM EMG (C) of one full ascending stride cycle for standard stair ascent (solid) and stair ascent with 20% extra body mass (dotted). The fascicular behaviour was analyzed from peak GM EMG until lift-off (horizontal bars indicate the analyzed phase per conditions). Vertical lines indicate foot-contact times (lift-off right (LO$_{right}$), touchdown right (TD$_{right}$) and lift-off left (LO$_{left}$)) for both conditions, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD$_{left}$). For clarity, standard deviations are omitted.

Discussion

The purpose of the present study was to investigate the influence of step-height and added body mass on the behaviour of GM muscle fascicles during stair ascent. We found an increase in fascicular shortening with increasing step-height and hardly any change in muscle fascicle behaviour with added body mass.

Step-height

By increasing step-height, the body must be elevated over a greater vertical distance. Results show, that the ankle and knee joint moments were increased to overcome this increasing height difference. These increases in joint moments agree well with values reported in an earlier study [Riener et al., 2002]. In contrast to this previous study, however, we found an additional knee moment peak around lift-off at higher step-heights. This difference may be explained by the greater step-heights tested in the present study. The elevation of the body seems to be a coordinated action of both the ankle and knee joints. The peak ankle plantarflexion moment (trailing leg – between 50 and 60% stride cycle) provides elevation to clear the foot from the step, and at that point, the knee joint moment of the opposite leg (leading leg – between 10 and 20% stride cycle) "pulls" the body to the next step. In this part of the stride, the knee extension moment increased with increased height. Hip extension also elevates the body in this phase of the movement [Protopapadaki et al., 2007; Riener et al., 2002]. It is known that the ankle joint moment can rise to ~85% of the maximum moment during normal stair ascent [Reeves et al., 2007]. With increasing step-height, the ankle joint moment needs to be further increased, even to approach the maximum joint moment. The increase in GM muscle activity shows that this muscle is, at least in part, responsible for the increase in ankle joint moment.

As expected, the GM muscle fascicles shortened more (and faster) at higher step-heights, and this was associated with a higher ankle joint moment and a larger range of motion. Increases in fascicular shortening with increased inclination have been found previously in level walking for humans [Lichtwark and Wilson, 2006] and running on an incline in wild turkeys [Gabaldon et al., 2004]. However, besides inter-species differences, stair ascent differs from level walking or walking on an incline in that there is no clear heel landing in stair ascent and the step length is constrained by the stair dimensions.

In a previous experiment (Chapter 3), we investigated the influence of gait velocity on GM muscle fascicle behaviour during stair negotiation. Similar to the present results with more shortening occurring during the lift-off phase as step-height increased, higher gait velocities required more GM muscle fascicle shortening. The peak ankle moment, however, did not increase any further with higher gait velocities, whereas in the present

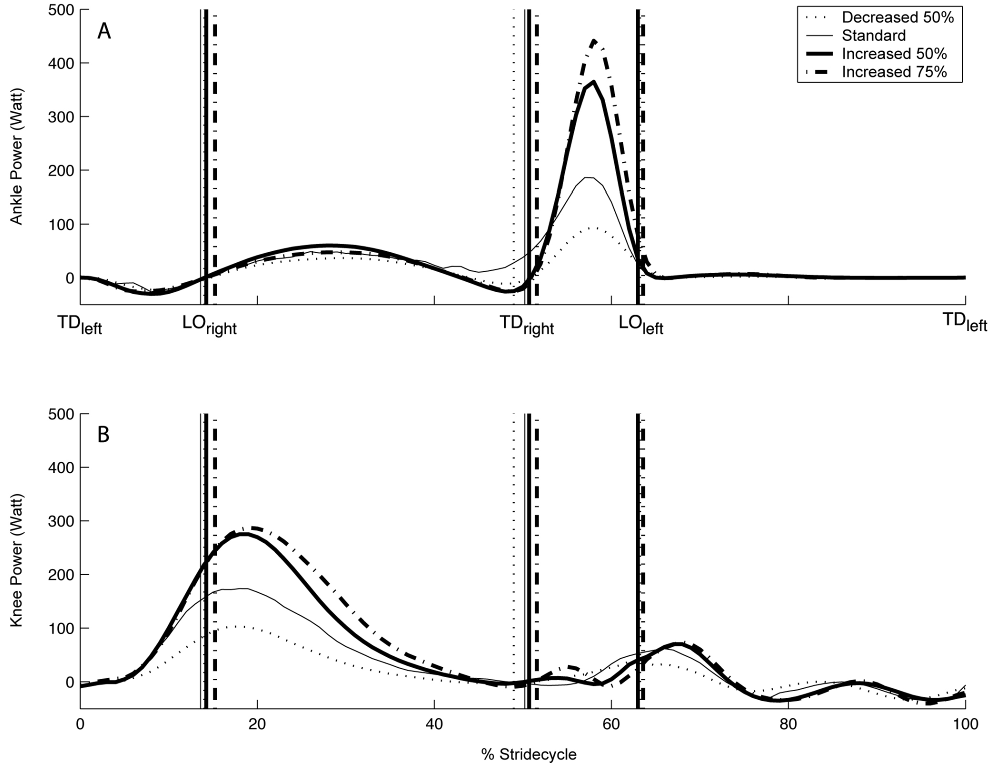

Figure 4.5. Ankle joint power (A) and knee joint power (B) of one full ascending stride cycle for all four step-heights: Decreased 50% (dotted), Standard (thin), Increased 50% (thick) and Increased 75% (thick and dashed). Vertical lines indicate foot-contact times (lift-off right (LO_{right}), touchdown right (TD_{right}) and lift-off left (LO_{left})) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD_{left}). For clarity, standard deviations are omitted.

study, the peak ankle moment increased far beyond the plateau observed in the earlier experiment. Probably because of this, and to keep the knee in an extended position, a knee moment peak was observed during lift-off that was not observed at normal step-heights and/or at higher ascending gait velocities. The overall increase in task demand is maybe best shown by the joint powers, which were calculated as the product of joint angular velocities and joint moments (Figure 4.5). During the lift-off phase, the peak in ankle joint power strongly increased. At this stage, the knee is extended and unable to produce much joint power. The increase in ankle joint power comes, at least in part, from the increased shortening of the GM muscle fascicles. However, it is impossible to measure the relative contribution of the GM muscle to the total increase in joint power non-invasively, since the GM muscle is not the only ankle plantarflexor muscle.

Added Mass

Extra mass caused the ankle and knee joint moments to increase during the whole stride cycle, except for the landing phase (first double support phase). It seems as though the extra mass is carried by the trailing leg until it lifts off. During the phase analyzed, there was no difference in EMG amplitude, which suggests that the GM muscle was activated similarly between trials, despite greater ankle joint moments with added mass.

The GM MTC length change was similar for the standard and added mass conditions, while the GM muscle fascicles were shorter for the added mass condition over the entire stride cycle. However, the amount of GM muscle fascicle shortening from peak EMG until lift-off did not differ between conditions, while the ankle joint moment did increase. So, a similar amount of fascicular shortening corresponded to a higher joint moment. To explain this surprising finding, we calculated the tendon length change. We estimated the total tendon length change (proximal and distal tendon and aponeurosis) as the difference between the MTC length change and the measured fascicle length change multiplied by the cosine of the pennation angle. The pennation angle was considered to be the angle that the fascicles made with the deep aponeurosis, which we measured from the same ultrasound images as the fascicle length [Kawakami et al., 1998; Spanjaard et al., 2007a]. The total tendon length change (Figure 4.6) was larger for the added body

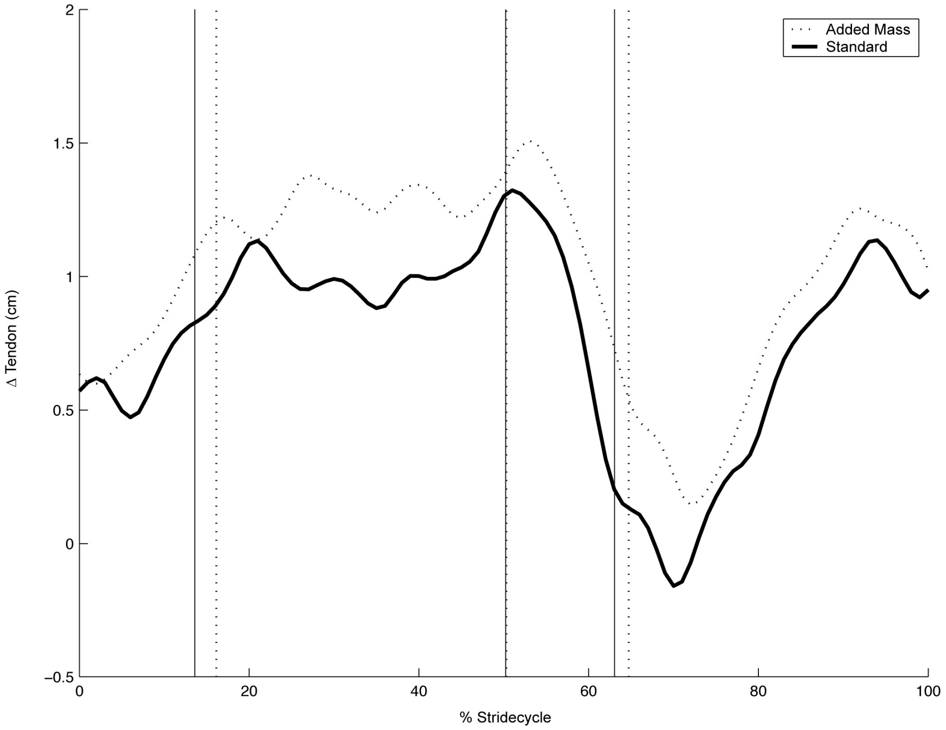

Figure 4.6. Mean tendon length change, relative to the tendon length calculated during standing still of one full ascending stride cycle for standard stair ascent (solid) and stair ascent with 20% extra body mass (dotted). Vertical lines indicate foot-contact times (lift-off right, touchdown right and lift-off left) both conditions, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot. For clarity, standard deviations are omitted.

mass trials over the whole stride cycle. This indicates that the tendon was stretched more under the influence of the extra mass. Muscle fascicle shortening would cause a higher joint moment when the tendon is stretched more. On the other hand, the shortening of GM muscle fascicles implies a left shift on the muscle force-length relationship [Gordon et al., 1966; Maganaris, 2003]. As this would be expected to lead to less force production by the fascicles for a given activation, the fact that the tendon was stretched more, while the EMG appeared unaffected, is interesting but difficult to explain. A possible explanation for the increase in tendon stretch and ankle joint moment in the absence of additional GM muscle activation may relate to the lower shortening velocity observed in added mass condition (although statistical significance was not reached, Table 4.2). Since at lower velocity the muscle has a higher force-generating potential [van Zandwijk et al., 1998] for a similar level of activation, a slower contraction will produce more force, more tendon stretch and a higher moment, consistent with the observations in the added mass condition. Another factor that could partly explain the differences in GM muscle fascicle length between conditions, especially when the ankle joint moment is much below its maximum value during the stride cycle, may relate to activation of parallel contractile elements within the triceps surae complex. If, for instance the gastrocnemius lateralis (GL) muscle activation increases more in the added mass condition, this muscle will pull harder on the Achilles tendon and the ankle plantarflexion moment will increase. Due to this tendon stretch, the GM muscle could shorten more without increasing its activation.

Changing the demands of stair ascent in different ways showed diverging effects on the kinematics and kinetics. Increases of step-height caused increased joint moments and increased GM muscle fascicle shortening during lift-off, whereas increase in body mass also showed increased joint moments with no further increase in GM muscle fascicle shortening during lift-off. The GM muscle fascicles were, however, shorter over the whole stride cycle with added body mass. In our previous study (Chapter 3) we found that higher gait velocity was associated with higher joint moments and increased GM muscle fascicle shortening during the lift-off phase of stair ascent. This pattern corresponds well with the results of the increased step-height, but contrasts with the results of the added body mass manipulation. The body mass manipulation trials were expected to show a similar pattern of fascicle length changes to the gait velocity manipulations because the dimensions of the staircase were the same in both conditions. However, increases in body mass resulted in no extra fascicle shortening during the analyzed phase of stair ascent. Thus, it appears difficult to predict the behaviour of muscle fascicles with changes in task demands.

Chapter 5

Lower-limb biomechanics during stair descent: Influence of step-height and body mass

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Abstract

The aim of the present study was to examine the biomechanics of the lower limb during stair descent and the effects of increasing demand in two ways: by increasing step-height and by increasing body mass. Ten male subjects walked down a four-step staircase, the height of which could be altered. The step-heights were: standard (17 cm), 50% decreased, 50% increased and 75% increased. At the standard height, subjects also walked down wearing a weighted jacket carrying 20% extra body mass. Lower limb kinematics and kinetics were determined using motion capture and ground reaction forces. Also measured were gastrocnemius medialis (GM) muscle electromyography and GM muscle fascicle length using ultrasonography. GM muscle fascicles actively shortened during the touch-down phase of stair descent in all conditions, while the musculotendon complex (MTC), as calculated from the knee and ankle joint kinematics, lengthened. The GM muscle fascicles shortened more when step-height was increased, which corresponded to the increase in ankle joint moment. Increased body mass did not alter the ankle or knee joint moment in the first contact phase of a step down; due to a change in strategy, the trailing leg, instead of the leading leg, supported the extra mass. Hence, the amount of GM muscle fascicle shortening, during the touch-down phase, did also not change with added body mass. Results suggest that the increase in joint moments is related to the amount of fascicle shortening, which occurs whilst the MTC is lengthening, thereby stretching the elastic tendinous tissues.

Introduction

Stair descent requires high lower-limb joint moments [McFadyen and Winter, 1988; Protopapadaki et al., 2007; Reeves et al., 2008; Riener et al., 2002], but instead of shortening to provide propulsion as in level walking and stair ascent for example [Fukunaga et al., 2001; Lichtwark and Wilson, 2006; McFadyen and Winter, 1988; Spanjaard et al., 2007a; Spanjaard et al., 2007b], the active muscles lengthen to absorb kinetic energy that is gained when descending a step [McFadyen and Winter, 1988; Spanjaard et al., 2007a; Spanjaard et al., 2007b]. During the landing phase of stair descent, the ankle joint moment is first to peak, followed shortly by a smaller knee joint moment peak [McFadyen and Winter, 1988; Protopapadaki et al., 2007; Riener et al., 2002; Spanjaard et al., 2007a; Spanjaard et al., 2007b]. The timing and magnitude of the ankle joint moment during the initial landing phase of stair descent highlights the important role of the plantarflexors during this task. Despite lengthening of the entire muscle-tendon complex (MTC) during the initial landing phase of stair descent, we have previously shown that the fascicles of the gastrocnemius medialis (GM) muscle actually shorten (Chapter 2). The energy stored in the tendon during this landing phase is then released at a later stage of the stride cycle stretching the muscle fascicles.

To understand further the role of the ankle joint in stair descent when real-world situations cause the task demands to alter, we recently manipulated gait velocity (Chapter 3). We investigated three gait velocities (63, 88 and 116 steps/min) during stair descent and we expected that joint angle patterns would have been unaffected because the step dimensions remained unaltered. In contrast, however, increases in gait velocity resulted not only in ankle joint moment increases, but also in changes in joint angle patterns. In addition, the GM muscle fascicles shortened more with increasing velocity. Moreover, it was interesting to note that the consistent incremental pattern of MTC lengthening and muscle fascicle shortening with increasing gait velocity was not paralleled by the ankle joint moment, which did not increase beyond 88 steps/min.

To gain further insight into the role of the ankle joint and GM muscle when the demands of stair-descent tasks are altered, in the present study we independently manipulated two different parameters during stair descent: step-height and body mass. The ankle joint and, in particular, the GM muscle are expected to be highly influenced by these alterations since these are expected to accommodate the change in force and negative work required [Maganaris et al., 2006; McFadyen and Winter, 1988; Protopapadaki et al., 2007; Riener et al., 2002; Spanjaard et al., 2007a]. The underlying rationale for these experimental manipulations was to mimic situations encountered habitually in the real world. For example, step-height varies dependent upon the location of the staircase, with typically higher step-heights in private dwellings and lower step-heights in public places [Roys, 2001]. A greater total body mass relative to the proportion of lean body mass is a typical characteristic of a number of different populations such as obese people, older adults and pregnant women.

Besides mimicking real world situations, changing task demands that alter energetic requirements can also give us insight into how the GM muscle (the fascicles of which are known to shorten during MTC stretch) copes with these requirements. When step-height or body mass increases, the requirement for negative work increases. The former increases the vertical distance that the centre of mass (CoM) has to travel, while the latter likely increases the force with which the CoM has to be decelerated. It is anticipated that the ankle joint moment and negative ankle joint power will increase with both increased step-height and body mass. For both increased demands, changes in GM muscle fascicle behaviour could vary between 2 extremes: 1 – Muscle fascicles will shorten more, performing more mechanical work, which will increase the amount of negative work to be performed by other structures even more. 2 – The muscle fascicles will shorten less or even lengthen, which has the advantage of acting at a more favourable position in the force-velocity relation [Hill, 1953], besides the advantage of performing less positive mechanical work where net negative work is needed. A disadvantage of the latter would be that the initial state of the muscle and tendon might not be stiff enough to effectively decelerate body mass, since the tendon could be close to slack length. Therefore, it is hypothesized that increasing the demands by increasing step-height or body mass involves more GM muscle fascicle shortening. It is further hypothesized that, to withstand the increase in demands, the ankle joint moment will increase.

Methods

Subjects

Ten healthy male subjects (age: 24.9 ± 3.2 years, height: 1.82 ± 0.06 m, mass: 79.9 ± 9.1 kg, leg length: 95.7 ± 4.7 cm) participated in the measurements. All gave their written informed consent for the study. The same subjects participated in previously reported studies (Chapter 2, 3 and 4) of our laboratory. Ethical approval was gained from the ethics committee of the Institute for Biomedical Research into Human Movement & Health at the Manchester Metropolitan University.

Measurements

The set-up has been described in detail in Chapter 2. Subjects were asked to descend a custom-built steel staircase of four steps, barefooted, in a step-over-step manner. The steps were independently mounted on the floor. The height of the steps could be altered by using individual, purpose-built metal frames underneath the steps. The tread and width of the steps were always the same: 280 mm x 900 mm. The riser was set at four different heights: standard height (170 mm), 50% decreased height (85 mm), 50% increased height (255 mm) and 75% increased height (297.5 mm). The dimensions of the standard height are most frequently encountered in semi-public places [Roys, 2001].

Kinetic data were collected by forceplates which were embedded in, and in front of, the staircase. Three force plates (Kistler 9286A, 27 x 52 cm) with built-in amplifiers were embedded in the first three steps (from the ground), and one force plate (Kistler 9253A, 40 by 60 cm) with an external amplifier (Kistler 9865C) was embedded in the floor, in front of the staircase.

A 9-camera VICON 612 system (VICON motion systems Ltd., Oxford UK) was used to capture kinematic data. Retro-reflective markers were placed on bony landmarks, directly on the skin, or on appropriate tight-fitting clothing using double-sided tape. In total, 34 markers were placed on the body according to the standard “plug-in-gait” model of the VICON system. From the marker coordinates, knee and ankle angles were calculated.

GM muscle fascicle behaviour was assessed in vivo by ultrasound scans (Aloka SSD-5000, Tokyo, Japan) recorded in real-time during the stair descent trials. A linear 7.5 MHz probe (UST-579T-7.5) with 60 mm field of view was tightly secured around the left lower leg in the mid-sagittal plane of the GM muscle with a custom-built fixation device. The fixation device was made of a plastic cast, molded to fit the general contour of the calf, with a window for the probe. The probe was held rigidly by the cast, which was securely fixed on the calf using Velcro straps. The experimenter supported the probe cable throughout scanning. Sampling rate was 22 samples/s and image resolution was 768 x 576 pixels. The ultrasound scanning was synchronized with the kinematic, kinetic and EMG data using an external trigger.

A Bagnoli EMG system (Delsys Inc. Boston, MA, USA) was used to record the electrical activity of the GM muscle of the left leg. The recording electrodes were placed proximal to the ultrasound scanning probe in the mid-sagittal plane of the muscle. The sampling rate of the EMG recordings was set at 2000 samples/s.

Protocol

Anthropometric measurements were taken for each subject to scale the generic human plug-in-gait model in the VICON software (Oxford Metrics Inc.). Subsequently, the markers and EMG electrodes were positioned and data collection was initiated.

The subjects were tested at four different step-height and two different body mass conditions. After practise trials and feeling comfortable with descending the staircase (at that specific height), subjects performed one stair descent at a predefined gait cadence, dictated by a metronome, which was set at 88 beats per minute (BPM) (which has previously been shown to closely match the self-selected cadence in these subjects (Chapter 3)). The measurement was repeated if the cadence of the subject did not correspond with the beats of the metronome (as observed by the experimenter). Furthermore, only at the standard step-height, the subjects performed a descent with added mass. Body mass was increased by 20% using a custom-made jacket of which the pockets were filled with lead-covered pieces of known mass. The jacket was secured tightly around the shoulders and waist of the subject without interfering with the rest of the equipment. The pieces of lead were placed such that the extra mass was distributed uniformly over the trunk. If the descending cadence corresponded well with the metronome, the trial was used for further analysis. In all trials, subjects stood still on top of the staircase (a platform) and started the trial with their right foot. The trial ended when the subject was on the ground, off the force-plate.

Data analysis

The first full stride cycle of the left leg was considered a steady-state stride cycle (from the first touch-down point of the left foot on the second step down, to the second touch-down point of the left foot on the floor), as was indicated by an earlier study [Andriacchi et al., 1980]. This steady state stride cycle was used for further analysis. Kinematics and kinetics in 3-D for the ankle and knee joints were calculated from marker positions and forceplate data, using VICON software. Only the sagittal plane information was used for further processing.

The GM muscle fascicle lengths were measured from the recorded ultrasonographic images. On each ultrasound image from the analyzed stride cycle, GM muscle fascicle length was measured manually using Matlab (The Mathworks, Inc, Natick, MA). Muscle fascicle length was measured using the assumption that the fascicular trajectory was linear. The fascicle length measured in a standing position was the reference length for each subject. To account for individual differences, the fascicle length change was calculated as the difference between the reference length and the measured fascicle length during the analyzed stride cycle.

The GM MTC length change (muscle plus free tendon and aponeurosis in both distal and proximal ends) was estimated using the equations by Menegaldo et al. [2004], taking the ankle and knee joint kinematics as input.

EMG signals were band-pass filtered (20-450 Hz) by the Delsys system, then rectified and smoothed (2nd order low pass 5 Hz bi-directional filter) using Matlab. EMG signals from previous studies [McFadyen and Winter, 1988; Reeves et al., 2008; Riener et al., 2002; Spanjaard et al., 2007a; Spanjaard et al., 2007b] and the present study revealed that the GM muscle was active during the touch-down phase. In terms of GM muscle fascicle behaviour, the phase of interest started with touch-down and ended when the muscle fascicles shortened maximally. For this phase, joint moment peaks, the amount of GM muscle fascicular shortening, fascicle shortening velocity, MTC length and the raw EMG root mean square (RMS) were calculated. Repeated measures ANOVA and Student's *t*-test were used to statistically analyze the influence of step-height and the influence of added body mass, respectively.

The position of the body CoM was calculated using a custom script in VICON "bodybuilder", which was based on the regression equations by Dempster [1955]. The script allowed adjustment for the added body mass condition, where the extra 20% body mass was ascribed to the trunk. Furthermore, an estimation of the net mechanical power produced by all muscles during stair descent was made and was coined "locomotory power". For this estimation, the human body was modelled as a single point-mass (the CoM), which was influenced by a single force (the ground reaction force). The locomotory power was calculated as the product of the ground reaction force and the CoM derivative. Ankle and knee joint power, calculated as the product of joint moment and joint angular velocity, were compared with the locomotory power.

Results

Kinetic, kinematic and EMG data were obtained for all ten subjects. GM fascicle length data of six subjects in the added mass trials were not available due to technical problems related to storage of the ultrasound images. This was also the case for one subject in the 75% increased height condition.

Step-height

There were no differences in step cadence between the step-height conditions ($p=0.671$). However, relative foot contact-times were affected by step-height; the stance phase became shorter at higher step-heights ($p<0.005$).

Main effects of step-height were found for peaks in ankle and knee joint moment. In agreement with our hypothesis, both increased with step-height (Figure 5.1 and Table 5.1). The relative timing of the ankle moment peak occurred earlier in time for higher step-heights ($p<0.001$).

Table 5.1. Investigated parameters for all four step heights.

	Step-height					
	Decreased 50%	Standard	Increased 50%	Increased 75%	p-value	Post-hoc ($p < 0.0167$)
Cadence (steps/min)	92.0 ± 4.5	91.6 ± 3.9	92.5 ± 4.2	91.0 ± 5.4	0.671	
1st peak knee moment (Nm)	37.1 ± 24.0	63.1 ± 22.0	81.9 ± 21.9	96.6 ± 30.6	0.001	
1st peak ankle moment (Nm)	83.0 ± 17.3	110.1 ± 26.0	141.6 ± 28.1	158.6 ± 35.8	0.001	-50 < standard < +50
GM fascicle shortening at touch-down (cm)	-1.60 ± 0.49	-2.36 ± 0.47	-2.55 ± 0.74	-2.66 ± 0.63	0.001	-50 < standard
Max GM fascicle shortening (cm)	-2.76 ± 0.50	-3.18 ± 0.31	-3.41 ± 0.69	-3.60 ± 0.62	0.006	
GM fascicle shortening velocity (cm/s)	7.4 ± 3.4	5.3 ± 2.0	6.6 ± 5.1	7.7 ± 3.4	0.384	
MTC length change (cm)	-0.4 ± 0.1	-0.6 ± 0.1	-0.8 ± 0.2	-0.8 ± 0.1	0.001	-50 < standard
GM EMG RMS (µV)	49 ± 1.9	97 ± 7.2	84 ± 4.5	116 ± 4.5	0.027	
Minimal locomotory power (Watt)	-295 ± 89	-515 ± 113	-800 ± 246	-1019 ± 406	0.001	-50 < standard < +50
Minimal ankle power (Watt)	-155 ± 42	-310 ± 91	-582 ± 168	-752 ± 234	0.001	-50 < standard < +50 < +75
Min ankle power relative to min locomotory power (%)	55 ± 11	60 ± 9	73 ± 9	77 ± 16	0.001	standard < +50

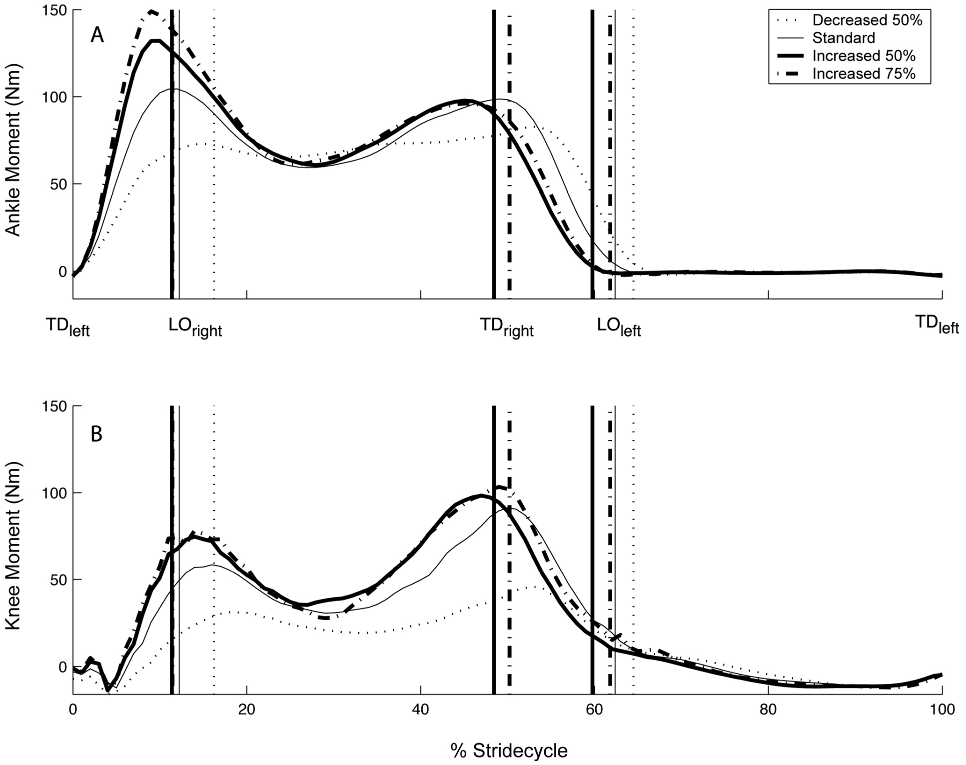

Figure 5.1. Mean ankle joint moment (A) and knee joint moment (B) of 10 subjects of one full descending stride cycle for all four step-heights: decreased 50% (dotted), standard (thin), increased 50% (thick) and increased 75% (dashed and thick). Vertical lines indicate foot-contact times (lift-off right (LO_{right}), touchdown right (TD_{right}) and lift-off left (LO_{left})) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD_{left}). For clarity, standard deviations are omitted. The standard deviation for the standard step-height can be viewed in Figure 5.5.

The locomotory power during a step down for all heights is shown in Figure 5.2. The maximal negative locomotory power, which occurred during the touchdown phase, increased with step-height (“minimal locomotory power” in Table 5.1). Exactly during this negative peak, the ankle joint power also showed a negative peak (Figure 5.3A), which also increased with step-height (Table 5.1). The relative contribution of the ankle power to the locomotory power during the negative power peak in touch-down increased with step-height (Table 5.1). Thus, the total amount of negative work on the whole body increased with step-height, as did the amount of negative work performed by the ankle plantarflexors.

GM muscle fascicle length change during a step down for all heights is shown in Figure 5.4A. In line with our hypothesis, the GM muscle fascicle shortening at touch-down increased with increasing step-height (Table 5.1). The timing (both relative and absolute) of maximal fascicle shortening did not differ between step-heights (p=0.503 and p=0.553,

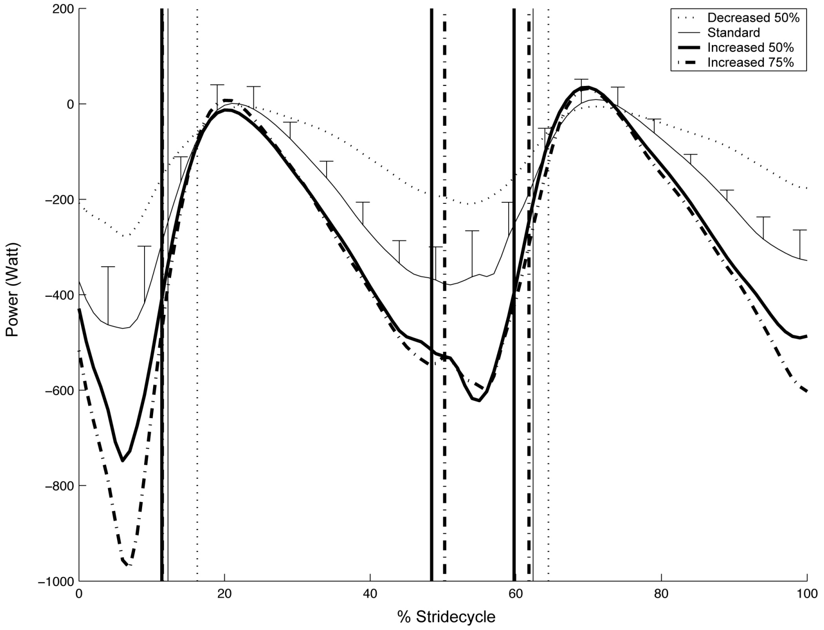

Figure 5.2. Mean locomotory power (the product of the ground reaction force and the centre of mass velocity) of 10 subjects of one full descending stride cycle for all four step-heights: decreased 50% (dotted), standard (thin), increased 50% (thick) and increased 75% (dashed and thick). Vertical lines indicate foot-contact times (lift-off right, touchdown right and lift-off left) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot. For clarity, positive standard deviations of only the standard step-height are presented.

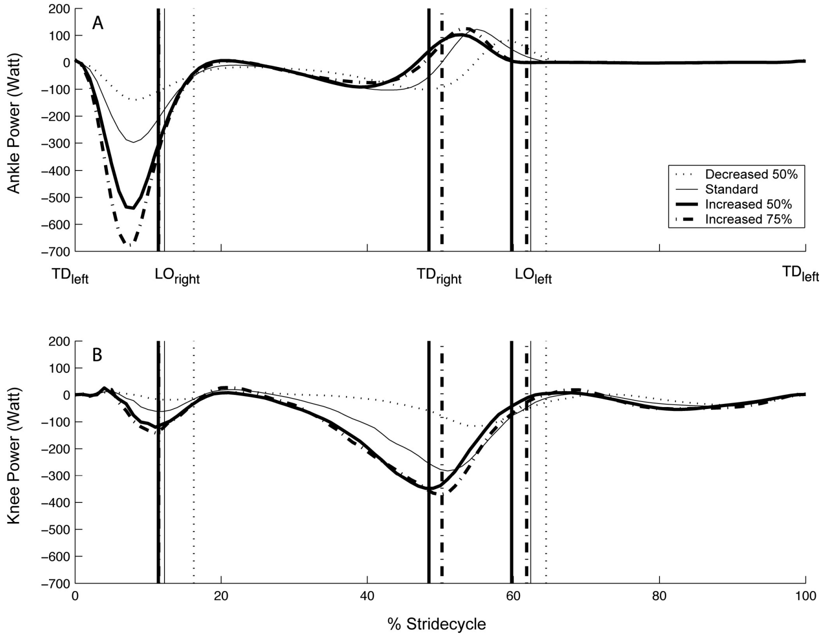

Figure 5.3. Mean ankle joint power (A) and knee joint power (B) of 10 subjects of one full descending stride cycle for all four step-heights: decreased 50% (dotted), standard (thin), increased 50% (thick) and increased 75% (dashed and thick). Vertical lines indicate foot-contact times (lift-off right (LO_{right}), touchdown right (TD_{right}) and lift-off left (LO_{left})) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD_{left}).. For clarity, standard deviations are omitted.

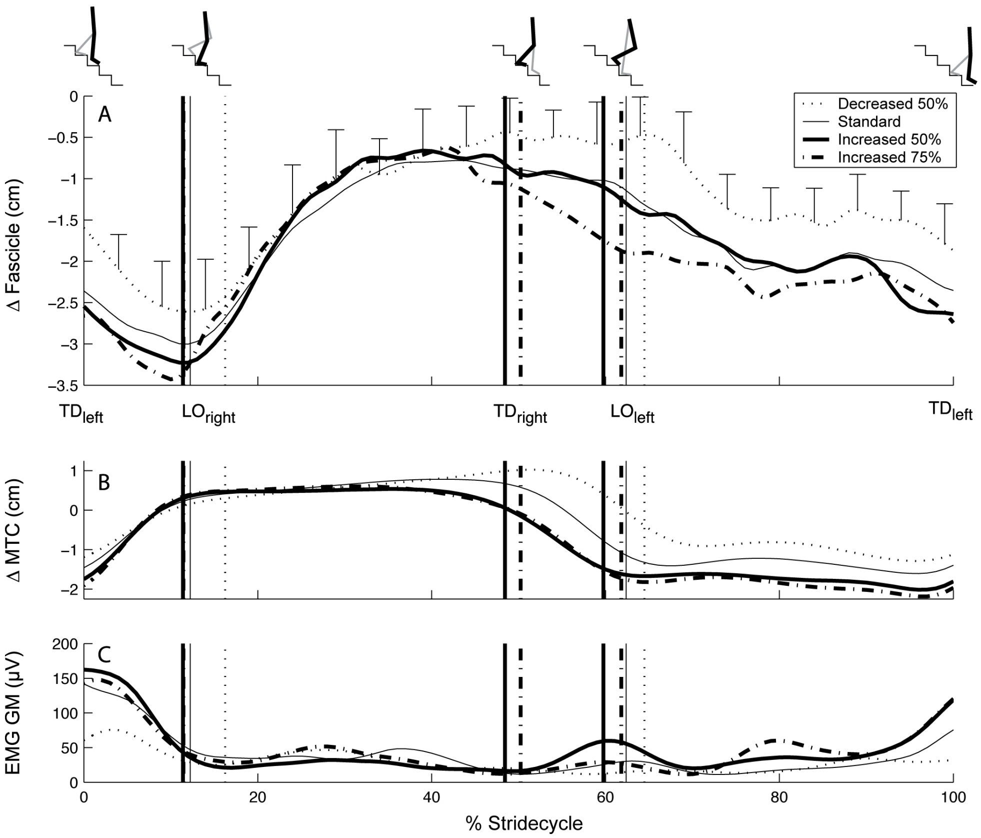

Figure 5.4. Mean fascicle length change (A), musculotendon complex (MTC) length change (B) and smoothed GM EMG (C) of 10 subjects of one full descending stride cycle for all four step-heights: decreased 50% (dotted), standard (thin), increased 50% (thick) and increased 75% (dashed and thick). The fascicular behaviour was analyzed during touch-down (from 0 to ~14% stridecycle). Vertical lines indicate foot-contact times (lift-off right (LO_{right}), touchdown right (TD_{right}) and lift-off left (LO_{left})) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD_{left}). Standard deviations were similar for all step-heights. For clarity, positive standard deviations of fascicle length change are plotted for the decreased 50% step-height only.

respectively) and no differences in shortening velocity were found between step-heights (Table 5.1). The reference fascicle lengths in the standing positions were: 6.54 ± 0.83 cm (decreased 50% height), 6.43 ± 0.54 cm (standard height), 6.46 ± 0.91 cm (increased 50% height) and 6.47 ± 0.90 cm (increased 75% height).

The kinematics were altered for different step-heights, as was shown by the MTC length (Figure 5.4B, Table 5.1), which was shorter for higher step-heights during the phase analyzed ($p<0.001$). This was due to more knee flexion over the entire stride cycle except for the point of first contact. The ankle joint had a larger range of motion at higher step-heights; more plantarflexion during touch-down and more dorsiflexion during lift-off.

Muscle activation was increased at higher step-heights, as indicated by the RMS values of the GM EMG data (Table 5.1).

Added Mass

There was no difference in step cadence between normal descent and descent with 20% added body mass (p=0.144). The first double support stance phase was longer in the added mass trials (p<0.005), while the duration of the total stance phase was not statistically different (p=0.068).

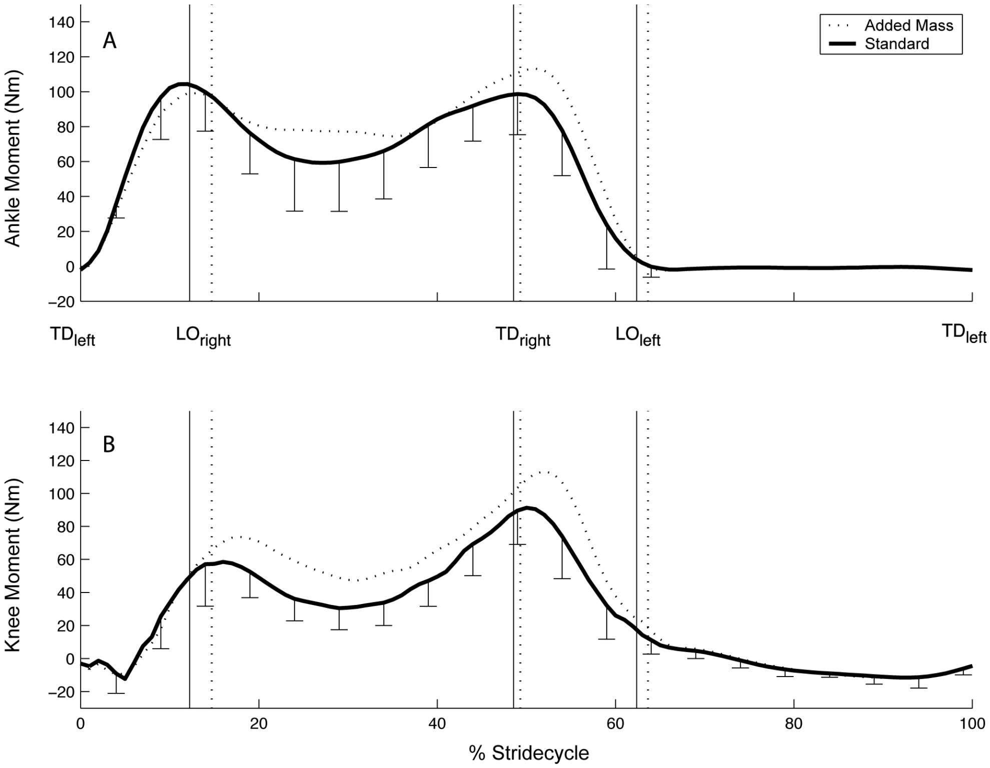

Figure 5.5. Mean ankle joint moment (A) and knee joint moment (B) of 10 subjects of one full descending stride cycle for standard stair descent (solid) and stair descent with 20% extra body mass (dotted). Vertical lines indicate foot-contact times (lift-off right (LO$_{right}$), touchdown right (TD$_{right}$) and lift-off left (LO$_{left}$)) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD$_{left}$). For clarity, only the negative standard deviations of standard stair mass condition are presented.

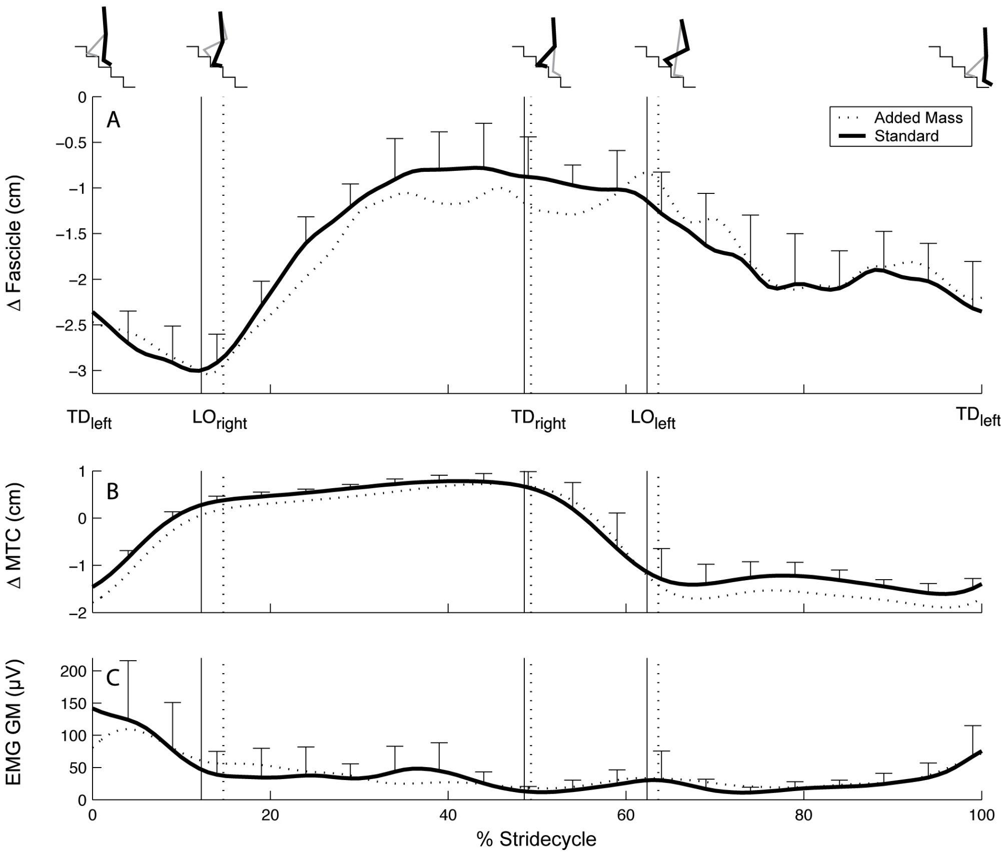

Figure 5.6. Mean fascicle length change (A) of 4 subjects, mean musculotendon complex (MTC) length change (B) and smoothed GM EMG (C) data of 10 subjects of one full descending stride cycle for standard stair descent (solid) and stair descent with 20% extra body mass (dotted). The fascicular behaviour was analyzed during touch-down (from 0 to ~14% stride cycle). Vertical lines indicate foot-contact times (lift-off right (LO$_{right}$), touchdown right (TD$_{right}$) and lift-off left (LO$_{left}$)) for all four step-heights, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot (TD$_{left}$). For clarity, only the positive standard deviations of the standard mass condition are presented.

In contrast to our hypothesis, the ankle joint moment did not change when body mass was increased during the touch-down phase (Figure 5.5A and Table 5.2). The timing of peak ankle moment did also not change with added mass (p=0.495). The peak knee joint moment during the touch-down phase was higher in the added mass condition, however, this occurred only after lift-off of the trailing leg (Figure 5.5B and Table 5.2). The joint moments of the trailing leg (the second ankle and knee joint moment in Figure 5.5A and B) did increase with added body mass (p=0.005 and p=0.008 for respectively the ankle and knee joint).

The locomotory power for the added body mass condition was similar as for the standard condition (hence not shown). The negative peaks of both the locomotory power and the ankle joint power during touch-down were not statistically different between conditions (Table 5.2). Also, the contribution of the ankle power on the locomotory power during the negative power peak did not change with added body mass (Table 5.2). Although the locomotory power was not higher during the touch-down phase for the added mass condition, it was observed that higher values were maintained for a longer period, so that the total amount of work performed on the CoM was increased for the added body mass condition.

The effect of added body mass on the GM muscle fascicle length change during a single stride is shown in Figure 5.6A and Table 5.2. The amount of GM muscle fascicle shortening and the timing of maximal shortening were not influenced by an increase in body mass (p=0.659 and p=0.769, n=4 respectively, Table 5.2). GM muscle fascicle shortening during touch-down did not differ between conditions. Fascicle shortening velocity was also not influenced by an increase in body mass (Table 5.2). MTC length, averaged over the analyzed period or at the time of maximal fascicle shortening, was not affected by an increase in body mass (p=0.332 and p=0.150 respectively, n=4, see Figure 5.6B and Table 5.2). This indicates that the kinematics were hardly affected, which was also confirmed after inspection of the ankle and knee joint angle traces (not shown). Added body mass did also not have an effect on the GM EMG during the analyzed phase, this was shown by the RMS values (Table 5.2).

Discussion

The purpose of the present study was to investigate the influence of step-height and added body mass on the biomechanics of lower-limb during stair descent. The results of the present study show an increase in fascicular shortening and ankle and knee joint moments with increasing step-height, while negative ankle joint power increased, which is in line with our hypothesis. However, in contrast to our hypothesis, there was no change in muscle fascicle behaviour or the ankle joint moment during the touch-down phase with added body mass. Instead, the walking strategy was altered in such a way that the extra load was carried by the trailing leg.

Step-height

The period in which the plantarflexor muscles are predominantly active during stair descent is touch-down. The toe hits the next step first, then, the heel touches down by a dorsiflexion movement of the ankle, during which the plantarflexor muscles are active to control this movement. Increased step-height, in stair descent, requires more negative work during touch-down. This is reflected in the negative locomotory power (Figure 5.2) and in the negative ankle and knee joint powers, which all increased with increasing step-height (Figure 5.3, Table 5.1). The ankle joint moment is the most important actuator to perform negative work since the contribution of the ankle joint to the locomotory power during touch-down (where the locomotory power is most negative) was 59% at the standard height and increased to 75% at the highest step-height (Table 5.1). The knee and ankle joint powers were consistent with previous studies [Devita et al., 2007; McFadyen and Winter, 1988; Riener et al., 2002]. The ankle joint moment peak occurred earlier in time for higher step-heights (Figure 5.1), corresponding with the shorter stance phase, emphasizing the need to increase negative power around the ankle with increasing step-height. We, and others, have argued before that the ankle joint plays a key role in stair descent [Reeves et al., 2008; Riener et al., 2002; Spanjaard et al., 2007a]; this is now supported by the large negative ankle joint power and its contribution to the locomotory power during touch-down. When step-height is increased, the relative contribution of the ankle joint to the locomotory power increases even further (Figure 5.2 and 5.3, Table 5.1).

The GM muscle is an important contributor to the ankle plantarflexion moment. GM muscle activation was increased for higher steps, consistent with an increase in the ankle joint moment. However, the ankle dorsi-flexed faster at higher step-heights, while the plantarflexion moment was increased, causing the ankle joint power to be more negative. The GM muscle fascicles shortened during this phase, performing mechanical work, while the total MTC lengthened, performing negative work. Earlier, studies from our lab also indicated shortening of fascicles during MTC lengthening while walking down stairs (Chapter 2 and 3). This behaviour of muscle fascicles is not shown when walking down a decline by humans or other species [Gabaldon et al., 2004; Lichtwark and Wilson, 2006]. Besides interspecies disparities, this difference is probably due to the toe landing in stair descent where there is a heel landing during decline walking. In species that walk on their toes (cats) it has been found that during level walking, the GM muscle fascicles also shorten while the MTC is lengthening [Griffiths, 1991]. Furthermore, the decline which was used in the experiments by Lichtwark and Wilson [2006] was very small (10%).

With increasing step-height, the MTC needs to perform more negative work in less time, as indicated by the locomotory power and the ankle joint power. While the MTC is performing negative work, the muscle fascicles are shortening and thus performing positive work. Furthermore, the amount of GM muscle fascicle shortening increased with step-height, which suggests that more positive work is performed, while more net negative work is needed. If the MTC length would be similar for all step-heights, this would mean that the tendon is stretched more for the highest steps. However, the MTC is shorter for higher steps, caused by the change in kinematics. When we calculate the tendon length change as the difference between the MTC length change and the fascicle

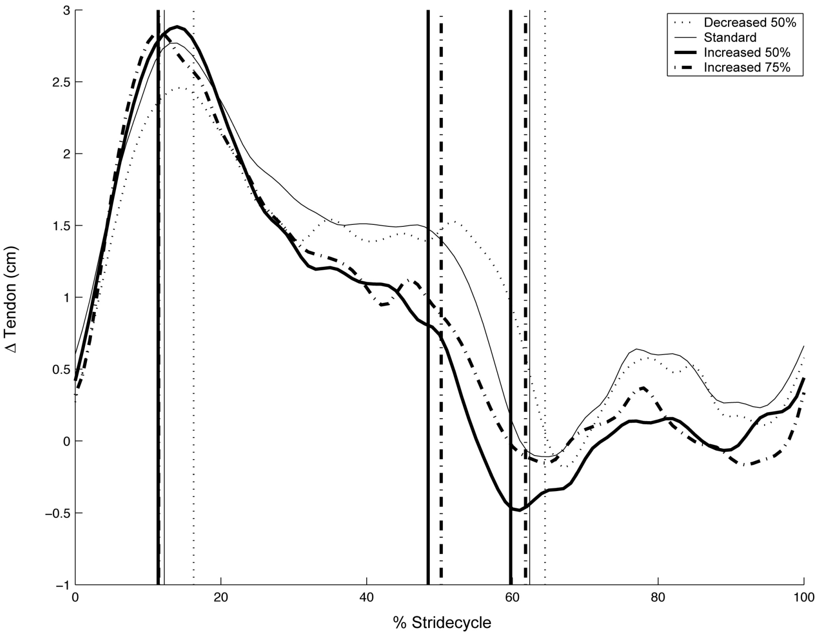

Figure 5.7. Mean tendon length change of 10 subjects, relative to the tendon length calculated during standing still, of one full descending stride cycle for all four step-heights: decreased 50% (dotted), standard (thin), increased 50% (thick) and increased 75% (dashed and thick). Vertical lines indicate foot-contact times (lift-off right, touch-down right and lift-off left) both conditions, whereas 0 and 100% of the stride cycle indicate touch-down of the left (analyzed) foot. For clarity, standard deviations are omitted.

length change, taking the pennation angle into account (the angle that the fascicles made with the deep aponeurosis, as was measured from the ultrasound images [Kawakami et al., 1993; Spanjaard et al., 2007a]), we find that the tendon is stretched more at higher step-heights, despite initial tendon lengths being similar (Figure 5.7). This suggests that the GM muscle fascicles produce more force at higher step-heights, as was also expected from ankle joint moment and EMG data, but they do this at shorter lengths, and thus probably further away from the optimal fascicle length [Gordon et al., 1966; Maganaris, 2003]. Presumably, the GM muscle fascicles need to shorten this much to increase the stiffness of the total MTC and control the dorsiflexion movement. Part of the kinetic energy from the touch-down and the extra mechanical work performed by the fascicle shortening appear to be stored in the tendon. Part of this energy is dissipated in mid-stance, while the other part of this energy seems to be used for ankle plantarflexion during lift-off.

In a previous experiment, we investigated the influence of gait velocity on GM muscle fascicle behaviour during stair descent (Chapter 3) and found an increase in the amount of fascicle shortening at higher gait velocities. This is similar to the results for the influence of step-height from the present study. However, the kinematics, and therefore the MTC lengths were quite similar between all velocities unlike in the present study for the different step-heights. Also, fascicle shortening velocity was increased for faster gaits, which was different than at higher steps, where we did not find any differences in fascicle shortening velocity. It seems that for both of the approaches employed to increase the task demands (increase in step-height and increase in gait velocity), the ankle joint moment and the GM muscle fascicle shortening were increased. Although the increase in shortening with increased joint moments does not seem unexpected, we stress that the GM muscle fascicles shortened in all conditions, while the MTC was lengthened by increasing external forces.

Added Mass

An increase of 20% body mass was expected to cause higher impact forces during the touch-down phase in the leg positioned on the step below. In contrast, however, the joint moments of the trailing leg increased (Figure 5.5, between 35 and 60% stride cycle). This suggests that to descend the stairs with added body mass a strategy was employed where the extra load was carried by the trailing leg. The leading leg was loaded relatively less (Figure 5.5, between 0 and 15% of stride cycle). The knee joint moment did increase, which occurred only after the touch-down phase when the trailing leg had just started its swing phase.

Because of the change in strategy with added mass, the plantarflexor muscles were not loaded more during touch-down with added mass than during normal stair descent, and the GM muscle fascicle behaviour was the same in both conditions.

General effects of increased demands on lower-limb mechanics

The shortening of the GM muscle fascicles during touch-down of stair descent controlled the dorsiflexion movement, while the MTC was lengthening. The GM muscle fascicles shortened more when step-height was increased, which corresponded with the increase in ankle joint moment. However, 20% extra body mass did not lead to extra shortening of the GM muscle fascicles. Due to a change in strategy, the mass on the leading leg was not increased, but instead, the trailing leg supported the extra mass. It would appear that these results of increasing the demands of stair descent would be predictable once ankle moments are known; higher ankle joint moments require more GM muscle fascicle shortening (increase in step-height), whereas when the ankle moment does not increase there is no extra GM muscle fascicle shortening (increased body mass). However, in a previous experiment in which we changed gait velocity to alter task demands (Chapter 3), we showed that when the peak ankle moment reaches a plateau and does not rise any further (with increasing gait velocity), the GM muscle fascicle shortening did increase further. This shows that the relation between fascicle shortening and ankle joint moment cannot be generalized to all situations where the task demands are altered. The non-linear elasticity of the tendon and the actions of other muscles are likely to play a role here. Also, the activation ratio between the three muscles of the triceps surae is known to change with movement velocity [Duchateau et al., 1986; Herman, 1967; Hof and van den Berg, 1977; Tamaki et al., 1997; Vandervoort and McComas, 1983], which may explain the difference between the influence of gait velocity and the influence of step-height on GM muscle fascicle behaviour. When the demands of the task were increased by adding body mass, the strategy was altered such that the leading leg, especially the ankle joint, was not loaded further. Therefore the amount of GM muscle fascicle shortening was similar between added mass and normal stair descent.

Methodological considerations

In the present study, the fascicular trajectory was approximated as a straight line, neglecting the slight curvature of the fascicles [Maganaris et al., 1998]. The difference between the two measurement approaches is, however, small (<3%, as estimated in Chapter 2) and falls within the observed variation (5.9%) of muscle fascicle length measurements performed by Narici et al. [1996]. The reliability of the muscle fascicle length measurements in the present study was calculated from the reference fascicle lengths which were measured during upright standing, on 4 different days. The intraclass correlation between these measurements was 0.8, indicating high reliability.

Although the probe was securely fixed on the skin, it is not known by how much the muscle did shift in relation to the scanning plane, therefore, we are unable to determine the precise magnitude of fascicle length measurement error introduced by scanning the muscle in 2-D. Data reported by Klimstra et al. [2007] indicate that a combined probe rotation by 5 degrees in the longitudinal direction and 5 degrees in the sagittal-frontal direction from the original scanning plane result in an average fascicle length error of no more than 8%. However, in the present study the ultrasound probe was securely fixed around the lower leg and no observable movement of the probe in relation to the leg could occur without application of external force to the probe manually. Moreover, it is highly unlikely that the GM muscle would rotated by as much as the experimental probe rotations examined by Klimstra et al. [2007]. Hence, we are confident that our fascicle length measurement errors introduced by 2-D scanning are much smaller than 8%. Errors in fascicle length measurements will be propagated in the calculation of fascicle shortening velocity.

To optimize image quality, the ultrasound sampling rate in the present study was set to 22 samples/s, which means that only 3.5 ultrasound frames were recorded in the phase of interest. We considered this compromise reasonable and we trust that our results and conclusions are valid since we were interested in general patterns of fascicle length changes. Furthermore, we believe that the frequency content of the fascicle length data is well below half the sampling frequency (11 Hz) [Bawa and Stein, 1976; Hidler et al., 2002].

The model by Menegaldo et al. [2004] which we used to calculate MTC length changes from ankle and knee joint kinematics as used in the present study show similar results as other models available. In fact, MTC length change values calculated according to the model by Menegaldo et al. [2004] are in between the values calculated according to the models by Grieve et al. [1978] and by Hawkins and Hull [1990]. The pattern of variation for MTC length is the same regardless of the model used. However we opted for using the model by Menegaldo and colleagues because this was based on detailed 3-D measurements of bone geometry and muscle-tendon origin and insertion positions from a living subject rather than cadavers.

Another point of discussion is the contribution of the different muscles in the triceps surae to the tendon stretch (as a measure of force applied) and the total joint moment. It is possible that a large part of the plantarflexion moment is produced by the soleus muscle (due to its large physiological cross sectional area). However, previous work has shown that the soleus muscle is already fully activated when the ankle joint moment reaches ~70% of maximal voluntary contraction (MVC) [Maganaris et al., 2006]. Any further increase in joint moment can thus be ascribed to both heads of the gastrocnemius muscle. Earlier studies also showed that the ankle joint moment during stair descent (at standard height) can increase up to ~75% of MVC [Reeves et al., 2008]. This means that any modulation around this peak in joint moment can be ascribed to the gastrocnemius muscle. Furthermore, the ankle joint moment increased even further at higher step-heights, it is expected that the relative contribution of the gastrocnemius muscle as compared to the soleus muscle will only increase in these circumstances. Nevertheless, we are presently unable to experimentally confirm this notion because we cannot measure the force contribution of different muscles converging in a common tendon.

Conclusion

The GM muscle fascicles shortened during the touch-down phase of stair descent while the GM MTC was lengthening. This indicates that the muscle fascicles were performing positive work while the whole GM MTC was performing negative work, consistent with the negative ankle joint power. With increased step-height, the requirements for net negative work of the lower extremity increased, which resulted in an increase in GM MTC lengthening, suggesting an increase in negative work performed by the GM MTC. In contrast, the GM muscle fascicles shortened more at higher step-heights, performing more positive work, which was consistent with our hypothesis. Adding 20% body mass altered the movement strategy in such a way that the extra load was supported by the trailing leg during touch-down, which resulted in the leading leg being loaded as in the normal situation. Therefore, GM muscle fascicle behaviour was not altered under the influence of extra body mass, which contradicted our hypothesis.

Chapter 6

Epilogue

Synthesis of results

The purpose of the present thesis was to gain insight into the modulation of gastrocnemius medialis (GM) contractile behaviour by the elasticity of the tendon in the complex locomotory task of stair negotiation in different demand conditions. The GM muscle fascicle behaviour was visualized in vivo with the use of ultrasonography, while joint kinematics and kinetics were also measured to fully understand the relationships between fascicle length changes, MTC length changes and joint moments exerted at the ankle and knee joints. In the studies described in the first experimental chapter (Chapter 2 in the thesis), subjects walked up and down a staircase of standard dimensions at a self selected velocity. After establishing the lower-limb biomechanics and behaviour of the GM muscle during stair negotiation under ‘standard’ conditions, the work moved on to investigate the influence of physiologically meaningful variations in the task demands. In Chapter 3, subjects performed stair ascents and descents at three predetermined step cadences on a staircase of standard dimensions. To analyze the contribution to positive work by the GM muscle in stair ascent in more detail, Chapter 4 was devoted to stair ascent only, where subjects walked up the stairs with variable step-heights, and with extra body mass attached to the trunk. In Chapter 5, the contribution to negative work by the GM muscle was analyzed in stair descent only by varying step-height and body mass. The results clearly show that muscle fascicle behaviour during stair negotiation does not correspond with MTC behaviour. During stair ascent, the GM muscle fascicles contracted near-isometrically during the lift-off phase, while the GM MTC shortened. In stair descent, the GM muscle fascicles shortened during the touch-down phase while the GM MTC lengthened. The extent of fascicle shortening was mainly related to the magnitude of joint moments. The increased demands imposed by increasing gait velocity, step-height and body mass resulted in increased ankle joint moments (except for the case of the highest cadence in Chapter 3), which were associated with an increase in fascicular shortening, even when the whole MTC lengthened.

During the lift-off phase of stair ascent (when the GM muscle was mainly active), the GM MTC shortened while the GM muscle fascicles contracted near-isometrically (with a tendency to shorten). During this phase – when the ankle joint moment peaked – the GM muscle fascicles were expected to shorten, since they would then contribute to the positive work, which is required for lifting the body upwards. So, the muscle fascicles did not contribute to work generation directly. However, by contracting near-isometrically the fascicles produced force at little metabolic cost, considering the force-velocity relation. The combination of fascicle force and external force on the foot caused the tendon to shorten (after an initial stretch), resulting in shortening of the whole MTC and a contribution of the MTC to the required ankle joint power. With increasing demands, which caused joint moments to increase, the GM muscle fascicle shortening increased. This suggests that the extra power needed to overcome the increased demand is, at least in part, produced by the GM muscle fascicles.

In the touch-down phase of stair descent the ankle joint moment peaked, which also caused a negative peak in ankle joint power. The knee joint moment showed its first peak shortly after the ankle joint moment peak and had a second (higher) peak just before lift-off. With increasing demands (step cadence and step-height), the ankle and knee joint moments increased. With an increase in body mass, the walking strategy changed, such that the trailing leg was loaded relatively more. This resulted in an ankle joint moment during touch-down similar to that during standard stair descent (no extra mass). After the touch-down phase, when the contra-lateral leg was in its swing phase, the ankle and knee joint moments increased due to the extra mass. During touch-down in stair descent, the GM muscle was most active and the ankle joint moment peaked. In this phase the GM MTC lengthened absorbing energy (thus performing negative work). However, the muscle fascicles shortened, performing positive mechanical work, while the task (lowering the body) required absorption of energy. Apparently, shortening was needed to bring the GM MTC into a state where energy absorption was possible (lengthening and producing force), and to provide enough ankle stiffness to ensure a safe descent. Increasing demands in stair descent, which caused the ankle joint moment to increase (increase in gait velocity and step-height), resulted in more GM muscle fascicle shortening. When gait velocity or step-height was increased, more energy had to be absorbed, while more energy was produced only by the extra shortening of the muscle fascicles.

Even though the isometric behaviour of the GM muscle fascicles during stair ascent was not expected (sonomicrometric data of running turkey’s have shown GM muscle fascicle shortening during uphill running [Gabaldon et al., 2004; Roberts et al., 1997]), the data of the present thesis resemble those of other in vivo studies focussed on stretch-shortening cycles. Ultrasonographic in vivo measurements during walking, running and jumping, all showed that the GM muscle fascicles contract mainly isometrically, producing high force levels, while the tendon stores and re-uses energy (providing the necessary work) [Fukashiro et al., 2006; Fukunaga et al., 2001; Ishikawa et al., 2005a; Ishikawa et al., 2007; Kawakami et al., 2002; Kawakami and Fukunaga, 2006; Kurokawa et al., 2001; Kurokawa et al., 2003; Lichtwark and Wilson, 2006]. The fascicle behaviour during a ‘standard’ ascent was considered to be isometric, since statistical analysis showed that the amount of shortening (there was a tendency to shorten) was not significantly different from zero. However, when the demands of stair ascent increased and more energy production was required, the GM muscle fascicle shortening increased. This resembles the results of turkey's running uphill with increased demands (faster uphill running – more GM muscle fascicle shortening) [Gabaldon et al., 2004].

The shortening of GM muscle fascicles during MTC lengthening in stair descent was unexpected. However, this paradoxical muscle movement has been reported before. Ishikawa et al. [2005b] found GM muscle fascicle shortening during GM MTC lengthening in short contact high-intensity drop jumps, which suggested that this behaviour might be seen during short, intense eccentric actions of muscles with (probably) long tendons. However, fascicle shortening during MTC lengthening has also been found in rat muscles (in-situ experiments) and in freely walking cats [Ettema et al., 1990; Griffiths, 1991].

Demands of stair negotiation

Earlier studies have indicated that stair negotiation is a strenuous task [Devita et al., 2007; McFadyen and Winter, 1988; Riener et al., 2002; Stacoff et al., 2005]. Compared to level walking, the joint ranges of motion, the ground reaction forces and the joint moments are larger [McFadyen and Winter, 1988; Riener et al., 2002; Stacoff et al., 2005]. The amount of work that muscles have to perform (positive or negative, for ascent or descent, respectively) is much higher during stair negotiation than during level walking [Devita et al., 2007]. As stated earlier, the joint moments during stair negotiation can reach values of 30-50% of MVC at the knee joint and 75-85% of MVC at the ankle joint [Reeves et al., 2007; Reeves et al., 2008]. The present thesis also indicates that stair negotiation is very demanding. The additional information that the ultrasound measurements provided, showed that fascicular behaviour was not similar to MTC behaviour, leading to high levels of tendon stretch. During ascent, the GM muscle fascicles contracted near-isometrically just before lift-off, which is an efficient type of contraction. However, during descent, the muscle fascicles shortened around the phase where the peak ankle joint moment took place. This implies that the muscle fascicles generating the high ankle joint moment were in a state of a relatively low force-producing potential according to the force-velocity relation [Hill, 1953]. Furthermore, it is known that the GM muscle acts on the ascending limb of the force-length relationship; hence, shorter fascicle lengths would be related to a reduced force-generating capacity [Gordon et al., 1966; Maganaris, 2003].

Increasing the demands of stair negotiation by increasing gait velocity, step-height and body mass (Chapters 3, 4 and 5, respectively) resulted in joint moment increases. The GM muscle fascicles shortened more, with a higher shortening velocity (increase in step-height) or with similar shortening velocities (increase in step cadence). So, the increased ankle joint moment (which came close to 100% of MVC) had to be generated with the GM muscle fascicles being in a state of even lower force-producing potential. These findings may partly explain why stair negotiation and especially stair descent is a demanding locomotory task.

Methodological considerations

The validity and reliability of the measurements with ultrasound in vivo have been critically assessed in Chapters 2 and 5, as well as in other studies [Kawakami et al., 1993; Loram et al., 2006; Maganaris, 2005; Narici et al., 1996]. It has been shown that the ultrasound measurements of fascicle length and pennation angle yield similar results as magnetic resonance imaging [Muramatsu et al., 2002] and direct anatomical inspection [Narici et al., 1996]. Furthermore, the repeatability of the tests has been shown to be satisfactory, with an average coefficient of variation of ~8% [Klimstra et al., 2007; Maganaris and Baltzopoulos, 1999; Narici et al., 1996], which is in line with the intraclass correlation of 0.8 reported in Chapter 5 in the present thesis. Manual measurements of muscle architecture on the same ultrasound scan by the same experimenter have previously yielded a coefficient of variation of ~3% [Maganaris and Baltzopoulos, 1999]. This shows that the ultrasound-based measurement of fascicle length in vivo is reliable.

The images obtained by ultrasonography are two-dimensional (2-D), so one has to take specific care to avoid projection errors. The fascicles and their plane of contraction should remain consistent with the plane of visualization throughout contraction, since probe rotation, relative to the plane of contraction, affects the measured muscle fascicle length [Klimstra et al., 2007]. During the measurements in the present study, the ultrasound probe was fixed rigidly to the leg by a custom-built fixation device and the probe cable was supported by the experimenter, to avoid probe movement relative to the skin and underlying muscle. However, it is uncertain how the muscle fascicles behave in 3-D during contraction. This could possibly lead to projection errors on the ultrasound image. However, the ultrasound images in the present thesis were of good quality and did not show prominent signs of muscle rotation within the ultrasound field of view.

The present thesis focuses on how muscle fascicle behaviour relates to joint kinematics and kinetics, thereby also making inferences on how muscle fascicle behaviour relates to joint work and power. However, to measure the amount of work that muscle fascicles perform, one must know the force produced by the fascicles in addition to their length change. Presently, it is not possible to measure the force generated by the GM muscle and its fascicles non-invasively. Therefore, it is not possible to quantify the contribution of the GM muscle fascicles to the total amount of work performed around the ankle joint. In the present studies, the GM muscle force was inferred from the levels of EMG and tendon stretch. This does not allow quantitative estimates of force or work production, however, muscle activation and the amount of tendon stretch can be compared between conditions. An alternative (but invasive) approach would be to measure the amount of force in the tendon, using tendon buckles or optic fibres [Arndt et al., 1998; Finni et al., 2003; Fukashiro et al., 1995; Gregor et al., 1991; Komi et al., 1987; Komi et al., 1996]. However, the GM muscle has a common tendon with both the GL and soleus muscles [Bojsen-Moller et al., 2004], the Achilles tendon. Therefore, the amount of force measured in the tendon would not be equal to the force generated by the GM muscle fascicles. We might circumvent this problem by subscribing a portion of the force measured in the tendon to each of the three muscles involved. Such estimates have been made by assuming that the contributions of the three heads of the triceps surae muscle are related to their physiological cross sectional areas (PCSA). However, this method is valid only if all three muscles are activated relatively equally, and it has already been demonstrated that this is not the case, since activation of muscles depends not only on force but also on the speed of contraction, which in turn is affected by the proportion of fast- and slow-twitch fibres in the muscle (the gastrocnemius muscle is a rather mixed-fibre muscle, “faster” than the predominantly type-I fibre soleus muscle) [Altenburg et al., 2007; Carpentier et al., 1996; Duchateau et al., 1986; Duysens et al., 1991; Hof and van den Berg, 1977; Tamaki et al., 1997; Vandervoort and McComas, 1983].

Future directions

The present thesis focused on how the GM muscle fascicles behave during stair negotiation. However, the GM muscle is certainly not the only muscle playing a role during this task. Furthermore, the GM muscle is not the only muscle producing an ankle plantarflexion moment. The soleus muscle is also an important plantarfexor and – unlike the GM muscle – it is a mono-articular muscle, only working around the ankle joint. In a pilot study performed on one subject, we measured soleus muscle fascicle behaviour during stair descent. Results are presented in Figure 6.1 and suggest that the soleus muscle fascicle behaviour is similar to the GM muscle fascicle behaviour. Further research is needed to confirm and generalize this finding.

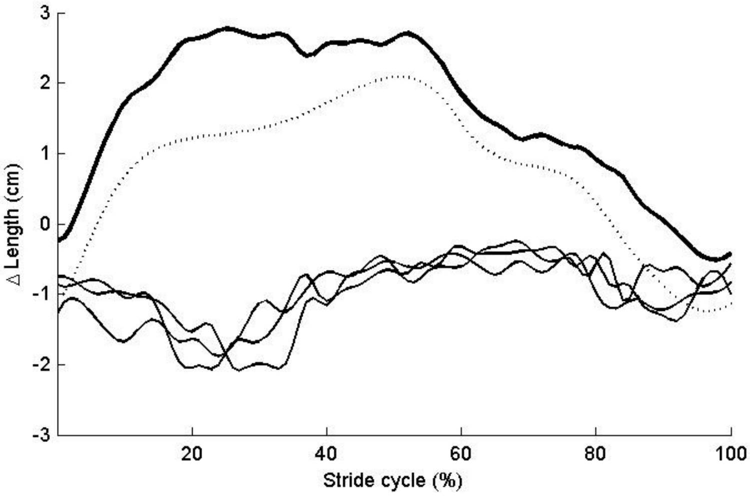

Figure 6.1. Individual data from three stair descending trials in a single subject showing soleus muscle fascicle length change (solid thin line), relative to the soleus muscle fascicle length measured during standing still. The soleus muscle fascicle length during standing still was 5.2 cm. Mean MTC length change data estimated from knee and ankle joint changes in 10 subjects (dotted line) and tendon length change data (solid thick line), are also presented.

The ankle joint provided a large part of the positive and negative work during ascent and descent, respectively (see for example Chapters 2 and 5). However, other joints, and thus other muscles, also acted to provide work. It is unlikely that the behaviour of the GM muscle fascicles is representative of the general functioning of all muscles involved in stair negotiation. Since the ankle plantarflexor muscles have a long and compliant tendon, the muscle fascicles will need to shorten more to attain the required stiffness. Fascicles from other muscles, such as the knee extensors, which have shorter and less compliant tendons, might not show the same behaviour since a short tendon would allow less fascicle shortening for a given force [Alexander and Ker, 1990].

Therefore, it is hypothesized that the fascicles of the knee extensors will shorten when the task requires positive work (stair ascent) and lengthen when the task requires negative work (stair descent). This hypothesis should be tested in future research.

The present thesis describes how muscle mechanics relates to whole body human biomechanics. However, muscle contractile length is largely determined by neural excitation and the nervous system also retrieves afferent signals on muscle length from the muscle spindles. Thus, by measuring contractile length directly (as in the present thesis), we have close access to information that the nervous system uses for controlling movement [Loram et al., 2006]. When muscle fascicle behaviour directly relates to whole MTC behaviour, the shortening and lengthening (and thus the efferent signals of the muscle spindles) of muscle fascicles might be predicted by joint kinematics. However, when muscle fascicle behaviour does not relate to whole MTC behaviour, as shown in the present studies, direct measurements of the contractile length are necessary to make inferences on motor control. This is clearly illustrated by the finding that the GM muscle fascicles shorten, while the MTC lengthens during stair descent. So, the muscle spindles of the GM muscle are probably not stretched during the touch-down phase of stair descent. Such effects have implications for motor control and should therefore be studied in more detail.

The relation between neural control, muscle behaviour and human movement is often studied with the use of computer models [Bobbert et al., 1990; Zajac, 1989; Zajac et al., 2002; Zajac et al., 2003]. In many models, muscles – as the actuators of movement – are controlled by stimuli (simulating those arising from the motor cortex or feedback loops), and cause the virtual body to move. The parameters of these computer models are partially based on physiological in vivo and in vitro studies of muscle and muscle fascicle geometry and function [Zajac, 1989]. For further validation of these parameters, ultrasonographic measurements can be used in future research.

The data of the present thesis might also be used for understanding the consequences of ageing and frailty for locomotion. It is known that the elderly have problems with stair negotiation [Cavanagh et al., 1997; Reeves et al., 2007; Reeves et al., 2008; Svanstrom, 1974; Tinetti, 2001]. There is strong evidence that this is related to an ageing-induced deterioration in contractile force-generating capacity [Reeves et al., 2007; Reeves et al., 2008]. This muscle functional deterioration is, to some degree, related to the changes in muscle architecture and tendon stiffness that occur with ageing [Narici et al., 2003]. The lower force generating-capacity in relation to the total body mass seen in the elderly (due to a loss in muscle mass), might be simulated in a young population by adding extra mass to the body, as done in Chapters 4 and 5 in this thesis. A step forward in understanding the consequences of ageing-induced changes in neuromuscular function for whole body locomotion would be made by directly investigating muscle fascicle behaviour in vivo in older populations. Comparisons of the present data with muscle fascicle behaviour data during stair negotiation in older people would give additional insight into the difficulties the elderly have to face in activities of daily life.

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Samenvatting

Spiervezelgedrag van de kuitspier tijdens traplopen

Spiervezels en peesweefsel vormen samen spierpeescomplexen, waarvan wij er heel veel in ons lichaam hebben. Spiervezels kunnen door middel van signalen uit het zenuwstelsel verkorten. Ze “trekken” dan aan het peesweefsel, dat vervolgens een kracht uitoefent op het bot, hetgeen vervolgens een beweging tot gevolg kan hebben. In het verleden zijn in onderzoek naar de eigenschappen en functies van verschillende spieren vaak uitspraken gedaan over spiervezelgedrag op basis van het verlengen of verkorten van het totale spierpeescomplex. Hierbij liet men echter de compliantie van de pees buiten beschouwing. Het is echter van belang om het gedrag van de spiervezels zelf te bestuderen, omdat deze de krachtleverende elementen van ons bewegingssysteem vormen.

In dit proefschrift wordt, met behulp van ultrasonografie, spiervezelgedrag (verandering van spiervezellengte) gemeten, om zo de relatie tussen spiervezel- en peesgedrag te beschrijven. Een “natuurlijke” en uitdagende taak om spiervezelgedrag te bestuderen is traplopen. Bij traplopen moeten de spierpeescomplexen van het been grote hoeveelheden arbeid (positief of negatief) genereren, veel meer dan bijvoorbeeld wandelen en is hierdoor het “schoolvoorbeeld” voor concentrische en excentrische contracties geworden. Bij traplopen moeten spierpeescomplexen verkorten en kracht leveren (concentrische spiercontractie) tijdens trap op lopen om de benodigde positieve arbeid te verrichten en ze (spierpeescomplexen) moeten verlengen en kracht leveren (excentrische spiercontractie) tijdens trap af lopen om de benodigde negatieve arbeid (remmen) te verrichten. In de experimenten die beschreven worden in dit proefschrift wordt spiervezelgedrag van de musculus gastrocnemius medialis (GM - binnenste oppervlakkige kuitspier) bestudeerd tijdens traplopen. Deze spier is een belangrijke plantairflexor van de enkel. Plantairflexie is het strekken van het enkelgewricht, een voor traplopen cruciale beweging waarmee veel van de benodigde arbeid tijdens traplopen wordt geleverd (zowel positief als negatief).

Na de introductie in hoofdstuk 1, beschrijft hoofdstuk 2 het eerste experiment. Tien proefpersonen liepen trap op en trap af op een zelf gekozen snelheid op een trap van 4 treden met standaard afmetingen. Hierbij werden de gewrichtshoeken (waaruit lengteveranderingen van het spierpeescomplex kunnen worden afgeleid) en gewrichtsmomenten van het enkel- en kniegewricht gemeten. Tevens werd de mate van spieractivatie van de GM met behulp van electromyografie gemeten. Tijdens traplopen werden ook de spiervezels in beeld gebracht door middel van ultrasonografie, waarbij de “ultrasound-probe” gefixeerd werd op het onderbeen van de proefpersoon. Achteraf werden de ultrasound beelden geanalyseerd en de spiervezellengtes gemeten. Bij trap op lopen was de GM voornamelijk actief tijdens de afzetfase. Tijdens deze afzetfase maakte de enkel een plantairflexie beweging, hetgeen een verkorting van het spierpeescomplex betekent. Echter, tijdens deze verkorting van het spierpeescomplex veranderden de spiervezels niet van lengte, maar contraheerde de spiervezels isometrisch, zij veranderden niet van lengte. Tijdens trap af lopen was de GM actief tijdens het neerkomen van de voet op de volgende trede. In deze fase verlengde het spierpeescomplex doordat de enkel een dorsiflexie beweging maakte. Echter, tijdens deze verlenging van het spierpeescomplex verkortten de spiervezels (!), wat er voor zorgde dat de pees sterk werd opgerekt. De lengte van de pees heeft een groot aandeel in de stijfheid van het spierpeescomplex, deze lengte wordt gereguleerd door de spiervezels. Blijkbaar vereist trap af lopen een hoge stijfheid van het spierpeescomplex en moeten de spiervezels dus verkorten. Deze resultaten laten zien dat het gedrag van spiervezels door de compliantie van het peesweefsel niet noodzakelijk hetzelfde is als dat van het spierpeescomplex. Deze resultaten zorgen er ook voor dat het gebruik van traplopen als “schoolvoorbeeld” van concentrische en (vooral) excentrische spiercontracties niet juist blijkt te zijn (voor deze specifieke spier, wel te verstaan).

Nadat het gedrag van de GM spiervezels tijdens “standaard” traplopen was vastgesteld werd gekeken naar de invloed van fysiologisch relevante variaties in de belasting van de taak. Hiertoe werden loopsnelheid, tredehoogte en lichaamsgewicht gemanipuleerd. Bij verhoging van de stapfrequentie (loopsnelheid) van traplopen (hoofdstuk 3), is er per stap een hoger mechanisch vermogen nodig (positief of negatief), simpelweg omdat dezelfde hoogte in een kortere tijd overbrugd moet worden. Dit vermogen moet geleverd worden door de spieren. Bij een hogere stapfrequentie vond meer GM spiervezelverkorting plaats, verkortten de spiervezels sneller en was het kniemoment hoger, voor zowel trap op als trap af lopen. Dit betekent dat tijdens trap af lopen op hogere snelheden (nog) meer energie geproduceerd werd door de spiervezels, terwijl er sneller energie geabsorbeerd diende te worden. Het enkelmoment werd ook hoger, echter tot een bepaalde stapfrequentie. Verdere verhoging van de stapfrequentie had geen verhoging van het enkelmoment tot gevolg terwijl de spiervezels wel verder verkortten. In hoofdstuk 4 en 5 werd de invloed van tredehoogte en lichaamsgewicht bestudeerd. Bij verhoging van tredehoogte moet er per stap meer mechanische arbeid worden verricht (positief of negatief), evenals bij een toename in lichaamsgewicht. Een verhoging van de traptreden had een verhoging van het enkelmoment tot gevolg, ook de GM spiervezels verkortten meer. Een verhoging van het lichaamsgewicht leidde echter tot andere resultaten. Tijdens trap op lopen verhoogde het enkelmoment wel, echter, de spiervezels van de GM verkortten niet meer vergeleken met de verkorting zonder extra lichaamsgewicht tijdens de afzetfase. In plaats daarvan waren de GM spiervezels tijdens de gehele stap korter. Tijdens de landingsfase van trap af lopen werd geen verandering van enkelmoment gevonden wanneer het lichaamsgewicht verhoogd was, in plaats daarvan compenseerde het achterste been voor de extra belasting. Dit had tot gevolg dat GM spiervezelgedrag tijdens de landingsfase niet werd beïnvloed door een verhoging van het lichaamsgewicht.

De belangrijkste bevinding van dit proefschrift is dat GM spiervezelgedrag niet noodzakelijk gelijk is aan het gedrag van het spierpeescomplex en dat GM spiervezelgedrag dus ook niet altijd nauwkeurig voorspeld kan worden op basis van het spierpeescomplex gedrag. Bij een verhoging van de belasting, die leidde tot een verhoging van gewrichtsmomenten werd meestal meer GM spiervezelverkorting gevonden, zelfs wanneer het spierpeescomplex verlengde (tijdens trap af lopen).

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