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

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SUMMARY

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 muscle–tendon 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, also did not change with added body mass. Our 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.

Key words: muscle mechanics, fascicle, ultrasound, musculotendon complex, stair descent.

INTRODUCTION

Stair descent requires high lower limb joint moments (McFadyen and Winter, 1988; Riener et al., 2002; Protopapadaki et al., 2007; Reeves et al., 2008), but instead of shortening to provide propulsion as in level walking and stair ascent for example (McFadyen and Winter, 1988; Fukunaga et al., 2001; Lichtwark and Wilson, 2006; 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 after by a smaller knee joint moment peak (McFadyen and Winter, 1988; Riener et al., 2002; Protopapadaki et al., 2007; 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 (Spanjaard et al., 2007a). The energy stored in the tendon during this landing phase was 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 (Spanjaard et al., 2007b). We investigated three gait velocities (63, 88 and 116 steps min^{-1}) 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^{-1}.

To gain further insight into the role of the ankle joint and GM muscle when the demands of stair-descent tasks are altered, 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 (McFadyen and Winter, 1988; Riener et al., 2002; Maganaris et al., 2006; Protopapadaki et al., 2007; Spanjaard et al., 2007a; Reeves et al., 2008). The underlying rationale for these experimental manipulations was to mimic situations encountered habitually in the real world. For example, step-height varies depending 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 two 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 additional advantage of acting at a more favourable position in the force–velocity relationship (Hill, 1953), besides that 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.

MATERIALS AND 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 had participated in previously reported studies (Spanjaard et al., 2007a; Spanjaard et al., 2007b) from our laboratory. Ethical approval was gained from the ethics committee of the Institute for Biomedical Research into Human Movement and Health at the Manchester Metropolitan University.

Measurements

The set-up has been described in detail previously (Spanjaard et al., 2007a). 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 using individual, purpose-built metal frames underneath the steps. The tread and width of the steps were always the same: 280 mm×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 force plates embedded in, and in front of, the staircase. Three force plates (Kistler 9286A, 27×52 cm) with built-in amplifiers were embedded in the first three steps (from the ground), and one force plate (Kistler 9253A, 40×60 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 joint 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, 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 throughout scanning. Sampling rate was 22 samples s^{-1} and image resolution was 768×576 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^{-1}.

Protocol

Anthropometric measurements were taken for each subject to scale the generic human plug-in-gait model in the VICON software (Oxford Metrics Inc., Oxford, UK). 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 min^{-1} [previously shown to closely match the self-selected cadence in these subjects (Spanjaard et al., 2007a)]. 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, and 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 with the pockets 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 weight 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 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 force plate 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, USA). 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 Menegaldo et al.'s equations (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; Riener et al., 2002; Spanjaard et al., 2007a; Spanjaard et al., 2007b; Reeves et al., 2008) 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 Dempster's regression equations (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 estimate of the net mechanical power produced by all muscles during stair descent was calculated and termed '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.

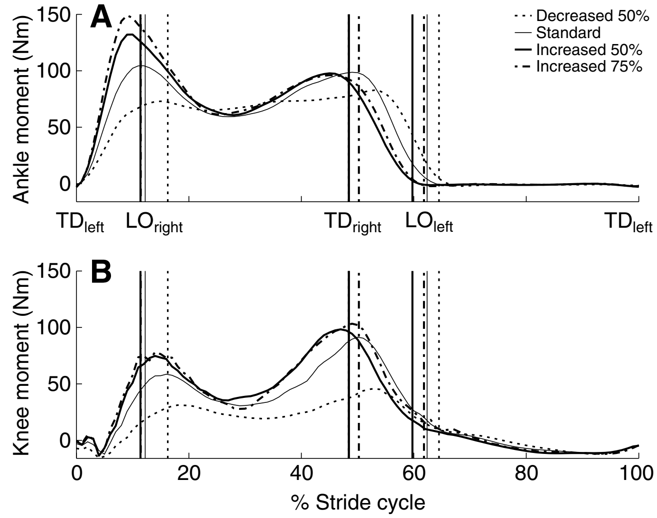

Fig. 1. Mean ankle joint moment (A) and knee joint moment (B) of 10 subjects of one full descending stride cycle for all four step-heights: decreased 50% (dotted lines), standard (thin lines), increased 50% (thick lines) and increased 75% (thick broken lines). Vertical lines indicate foot-contact times [lift-off (LO) right, touch-down (TD) right and lift-off left] for all four step-heights, whereas 0 and 100% of the stride cycle indicate TD of the left (analyzed) foot. For clarity, standard deviations (s.d.) are omitted. The s.d. for the standard step-height can be viewed in Fig. 5.

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).

The main effects of step-height were found for peaks in ankle and knee joint moments. In agreement with our hypothesis, both increased with step-height (Fig. 1 and Table 1). The relative timing of the ankle moment peak occurred earlier in time for higher step-heights (P<0.001).

The locomotory power during a step down for all heights is shown in Fig. 2. The maximal negative locomotory power, which occurred during the touchdown phase, increased with step-height ('minimal locomotory power' in Table 1). Exactly coincident with this negative peak, the ankle joint power also showed a negative peak (Fig. 3A), which also increased with step-height (Table 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 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 Fig. 4A. In line with our hypothesis, the GM muscle fascicle shortening at touch-down increased with increasing step-height (Table 1). The timing (both relative and absolute) of maximal fascicle shortening did not differ between step-heights (P=0.503 and P=0.553, respectively) and no differences in shortening velocity were found between step-heights (Table 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 (Fig. 4B, Table 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 plantar-flexion during touch-down and more dorsi-flexion during lift-off.

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

In contrast to our hypothesis, the ankle joint moment did not change when body mass was increased during the touch-down phase (Fig. 5A and Table 2). The timing of peak ankle moment did also not change with added mass (P=0.495). The peak knee joint

Table 1. Investigated parameters for all four step-heights

	Step-height					
	Decreased 50%	Standard	Increased 50%	Increased 75%	*P*-value	*Post hoc* analysis (P<0.0167)
Cadence (steps min^{-1})	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	
Ankle moment (Nm)	83.0±17.3	110.1±26.0	141.6±28.1	158.6±35.8	0.001	min50 < 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	min50 < 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^{-1})	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	min50 < standard
GM EMG RMS (μV)	49±1.9	97±7.2	84±4.5	116±4.5	0.027	
Minimal locomotory power (W)	–295±89	–515±113	–800±246	–1019±406	0.001	min50 < standard < +50
Minimal ankle power (W)	–155±42	–310±91	–582±168	–752±234	0.001	min50 < standard < +50 < +75
Min ankle power relative to min locomotory power (%)	55±11	60±9	73±9	77±16	0.001	standard < +50

P value refers to repeated measures ANOVA. Min50, step-height decreased by 50%; +50, +75, step-height increased by 50% or 75%, respectively.

moment during the touch-down phase was higher in the added mass condition, however, this occurred only after lift-off of the trailing leg (Fig. 5B and Table 2). The joint moments of the trailing leg (the second ankle and knee joint moment in Fig. 5A,B) did increase with added body mass (P=0.005 and P=0.008 for the ankle and knee joint, respectively).

The locomotory power for the added body mass condition was similar to that of 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 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 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 Fig. 6A and Table 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, respectively, N=4; Table 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 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 Fig. 6B and Table 2). This indicates that the kinematics were hardly affected, which was confirmed after inspection of the ankle and knee joint angle traces (not shown).

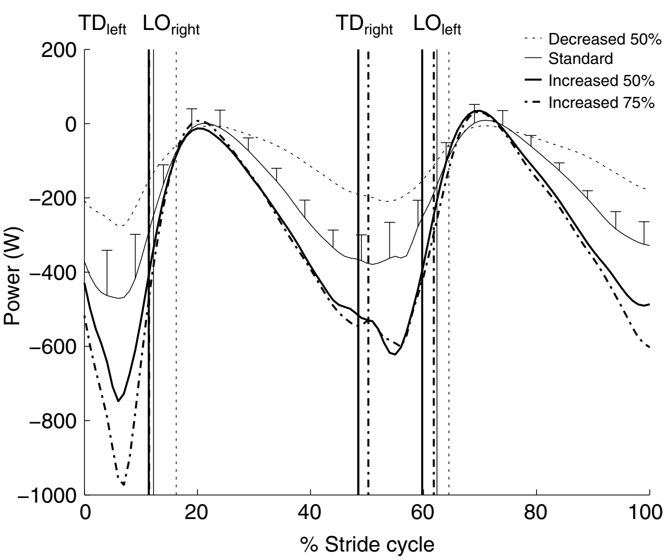

Fig. 2. Mean locomotory power (ground reaction force × centre of mass velocity) of 10 subjects over one full descending stride cycle for all four step-heights: decreased 50% (dotted lines), standard (thin lines), increased 50% (thick lines) and increased 75% (thick broken lines). Vertical lines indicate foot-contact times [lift-off (LO) right, touch-down (TD) right and LO left] for both conditions, whereas 0 and 100% of the stride cycle indicate TD of the left (analyzed) foot. For clarity, only positive standard deviations of the standard step-height are presented.

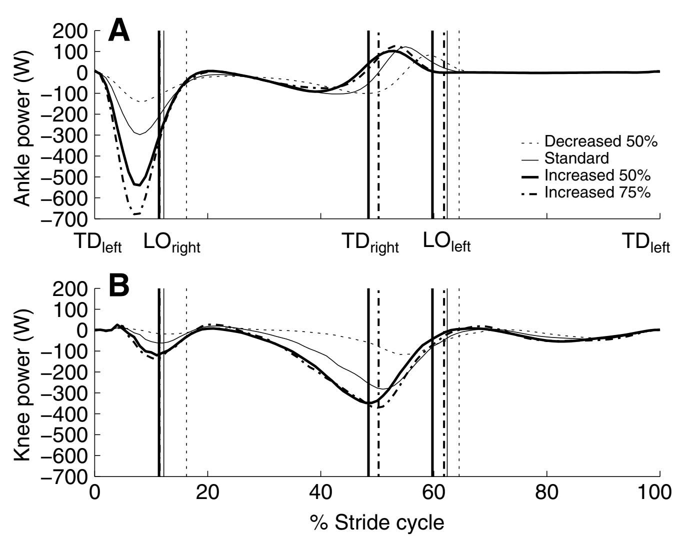

Fig. 3. Mean ankle joint power (A) and knee joint power (B) of 10 subjects of one full descending stride cycle for all four step-heights: decreased 50% (dotted lines), standard (thin lines), increased 50% (thick lines) and increased 75% (thick broken lines). Vertical lines indicate foot-contact times [lift-off (LO) right, touch-down (TD) right and LO left] for all four step-heights, whereas 0 and 100% of the stride cycle indicate TD of the left (analyzed) foot. For clarity, standard deviations are omitted.

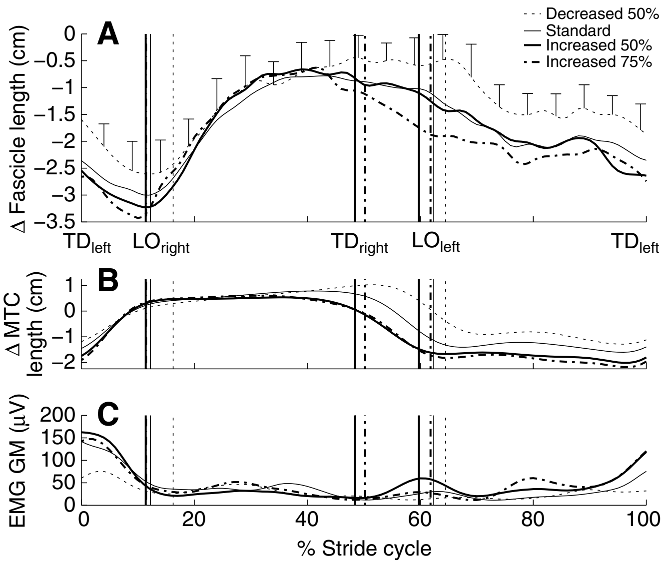

Fig. 4. Mean fascicle length change (A), muscle–tendon complex (MTC) length change (B) and smoothed GM EMG (C) of 10 subjects of one full descending stride cycle for all four step-heights: decreased 50% (dotted lines), standard (thin lines), increased 50% (thick lines) and increased 75% (thick broken lines). The fascicular behaviour was analyzed during touch-down (from 0 to ~14% stride cycle). Vertical lines indicate foot-contact times [lift-off (LO) right, touch-down (TD) right and LO left] for all four step-heights, whereas 0 and 100% of the stride cycle indicate TD of the left (analyzed) foot. Standard deviations (s.d.) were similar for all step-heights. For clarity, only positive s.d. of fascicle length change are plotted for the decreased 50% step-height.

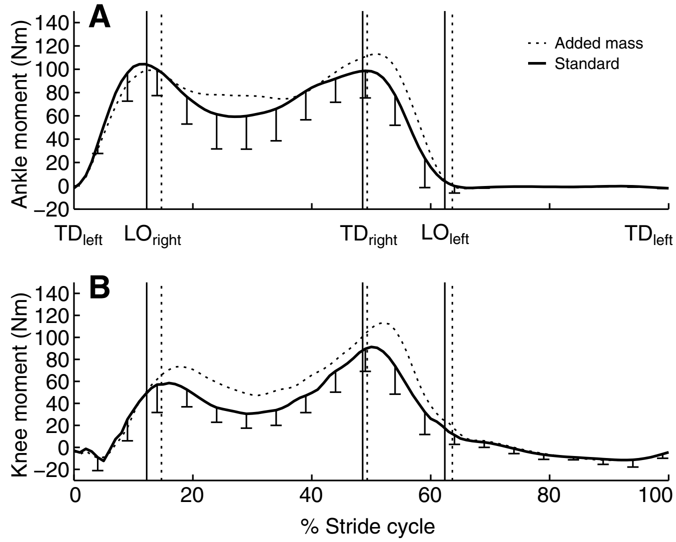

Fig. 5. Mean ankle joint moment (A) and knee joint moment (B) of 10 subjects of one full descending stride cycle for standard stair descent (solid lines) and stair descent with 20% extra body weight (dotted lines). Vertical lines indicate foot-contact times [lift-off (LO) right, touch-down (TD) right and LO left] for both conditions, whereas 0 and 100% of the stride cycle indicate TD of the left (analyzed) foot. For clarity, only the negative s.d. of standard stair mass condition are presented.

Added body mass did also not have an effect on the GM EMG during the analyzed phase, as shown by the RMS values (Table 2).

DISCUSSION

The purpose of the present study was to investigate the influence of step-height and added body mass on the biomechanics of the 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 in 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 dorsi-flexion 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 (Fig. 2) and in the negative ankle and knee joint powers, which all increased with increasing step-height (Fig. 3, Table 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 1).

Table 2. Investigated parameters for standard stair descent and stair descent with 20% extra body mass

	Body mass			
	Standard	Added 20%	*P*-value	*N*
Cadence (steps min^{-1})	91.6±4	89.9±4	0.144	10
1st peak				
Knee moment (Nm)	63.1±22.0	76.4±25.8	0.008	10
Ankle moment (Nm)	110.1±26.0	108.0±24.6	0.760	10
GM fascicle shortening at touch-down (cm)	–2.50±0.43	–2.47±0.27	0.912	4
GM fascicle shortening (cm)	–3.22±0.43	–3.11±0.64	0.659	4
GM fascicle shortening velocity (cm s^{-1})	4.5±1.7	4.2±2.9	0.674	4
MTC length change (cm)	–0.6±0.1	–0.8±0.2	0.332	4
GM EMG RMS (μV)	63±34	59±22	0.559	4
Minimal locomotory power (W)	–515±113	–545±152	0.450	10
Minimal ankle power (W)	–310±91	–284±89	0.235	10
Min ankle power relative to min locomotory power (%)	60±9	52±9	0.072	10

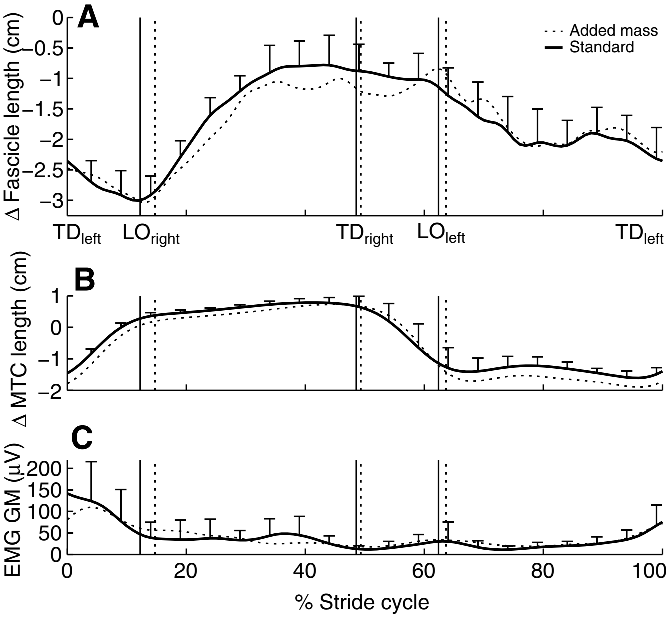

Fig. 6. Mean fascicle length change (A) of 4 subjects, mean muscle–tendon complex (MTC) length change (B) and smoothed GM EMG (C) data of 10 subjects of one full descending stride cycle for standard stair descent (solid lines) and stair descent with 20% extra body weight (dotted lines). The fascicular behaviour was analyzed during touch-down (from 0 to ~14% stride cycle). Vertical lines indicate foot-contact times [lift-off (LO) right, touch-down (TD) right and LO left] for both conditions, whereas 0 and 100% of the stride cycle indicate TD of the left (analyzed) foot. For clarity, only the positive standard deviations of the standard mass condition are presented.

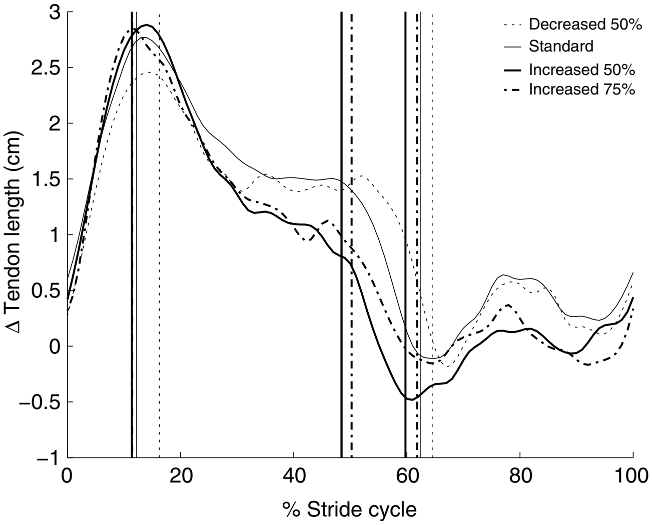

Fig. 7. Mean tendon length change of 10 subjects, relative to the tendon length calculated when standing still, of one full descending stride cycle for all four step-heights: decreased 50% (dotted lines), standard (thin lines), increased 50% (thick lines) and increased 75% (thick broken lines). Vertical lines indicate foot-contact times [lift-off (LO) right, touch-down (TD) right and LO left] for both conditions, whereas 0 and 100% of the stride cycle indicate TD of the left (analyzed) foot. For clarity, standard deviations are omitted.

The knee and ankle joint powers were consistent with previous studies (McFadyen and Winter, 1988; Riener et al., 2002; Devita et al., 2007). The ankle joint moment peak occurred earlier in time for higher step-heights (Fig. 1), corresponding with the shorter stance phase, and 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 (Riener et al., 2002; Spanjaard et al., 2007a; Reeves et al., 2008); 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 (Figs 3 and 5, Table 1).

The GM muscle is an important contributor to the ankle plantar flexion 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 plantar flexion 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 laboratory also indicated shortening of fascicles during MTC lengthening while walking down stairs (Spanjaard et al., 2007a; Spanjaard et al., 2007b). This behaviour of muscle fascicles is not shown when humans or other species walk down a decline (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 used in the experiments by Lichtwark and Wilson (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 increases with step-height, which suggests that more positive work is performed, while more net negative work is needed. If the MTC length were similar for all step-heights, the tendon would be 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 length change, taking into account the pennation angle [the angle that the fascicles made with the deep aponeurosis, as was measured from 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 (Fig. 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 dorsi-flexion 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

(Spanjaard et al., 2007b) 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 weight 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 (Fig. 5, between 35 and 60% stride cycle). This suggests that to descend the stairs with added body mass a strategy was employed whereby the extra load was carried by the trailing leg. The leading leg was loaded relatively less (Fig. 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 dorsi-flexion 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 weight on the leading leg was not increased, but instead, the trailing leg supported the extra weight. 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 (Spanjaard et al., 2007b), 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 (Herman, 1967; Hof and van den Berg, 1977; Vandervoort and McComas, 1983; Duchateau et al., 1986; Tamaki et al., 1997), 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 a previous experiment (Spanjaard et al., 2007a)] and falls within the observed variation (5.9%) of muscle fascicle length measurements performed by Narici et al. (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 shifted 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 elsewhere (Klimstra et al., 2007) indicate that a combined probe rotation by 5° in the longitudinal direction and 5° in the sagittal-frontal direction from the original scanning plane results 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 manual application of external force to the probe. Moreover, it is highly unlikely that the GM muscle would rotated by as much as the experimental probe rotations examined by Klimstra et al. (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^{-1}, 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. (Menegaldo et al., 2004) used to calculate MTC length changes from ankle and knee joint kinematics in the present study shows similar results to those obtained using other models available. In fact, MTC length change values calculated according to Menegaldo et al.'s model (Menegaldo et al., 2004) are in between the values calculated using other models (Grieve et al., 1978; Hawkins and Hull, 1990). The pattern of variation for MTC length is the same regardless of the model used. However, we opted for 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, so 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 confirm this notion experimentally 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, in contradiction to our hypothesis.

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