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Stair ascent and descent at different inclinations

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Abstract

The aim of this study was to investigate the biomechanics and motor co-ordination in humans during stair climbing at different inclinations. Ten normal subjects ascended and descended a five-step staircase at three different inclinations (24°, 30°, 42°). Three steps were instrumented with force sensors and provided 6 dof ground reactions. Kinematics was analysed by a camera-based optoelectronic system. An inverse dynamics approach was applied to compute joint moments and powers. The different kinematic and kinetic patterns of stair ascent and descent were analysed and compared to level walking patterns. Temporal gait cycle parameters and ground reactions were not significantly affected by staircase inclination. Joint angles and moments showed a relatively low but significant dependency on the inclination. A large influence was observed in joint powers. This can be related to the varying amount of potential energy that has to be produced (during ascent) or absorbed (during descent) by the muscles. The kinematics and kinetics of staircase walking differ considerably from level walking. Interestingly, no definite signs could be found indicating that there is an adaptation or shift in the motor patterns when moving from level to stair walking. This can be clearly seen in the foot placement: compared to level walking, the forefoot strikes the ground first—independent from climbing direction and inclination. This and further findings suggest that there is a certain inclination angle or angular range where subjects do switch between a level walking and a stair walking gait pattern.

Keywords: Stair climbing; Ascent; Descent; Motion analysis; Kinematics; Kinetics; Ground reactions; Stair inclination; Slope

1. Introduction

Stairs are frequently encountered obstacles in daily living. Although healthy persons climb stairs quite easily, this movement task is quite demanding when motor functions are reduced, for example in elderly or obese subjects, women during pregnancy, subjects affected by muscle or joint diseases as well as subjects with joint or limb replacements. The analysis of biomechanical and motor control aspects involved in stair ascent and descent can add to our understanding of the diverse and complicated processes involved in human locomotion and also be useful in the design of private and public environments where stairs are employed. Another application is in the field of gait rehabilitation. For instance, a comprehensive movement analysis of stair climbing can support the evaluation of joint replacement or prostheses development.

Several studies were performed to investigate normal human stair ascent and descent [1–3]. Recently, some researchers focused on the analysis of joint moments [4], joint powers [5], plantar pressure characteristics [6] and reproducibility [7] that occur during staircase walking. Some studies also exist that investigate stair climbing of patients with knee [8] and hip [9] implants, amputees with artificial limbs [10,11] or athletes with anterior cruciate ligament deficiencies [12].

However, no comprehensive analysis is available in the literature that discusses biomechanics of stair ascent and descent at different inclinations, although staircase slope proves to be an important characteristic affecting temporal and kinematic gait parameters [13] and hip/knee extensor activity [14] at varying step heights. Since previous studies [13,14] are restricted to rather specific issues, our study is a broader attempt to face the question of how staircase inclination affects the kine-

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matic and kinetic patterns of stair climbing and to ascertain if ascent and descent patterns are to be considered as particular evolution of the level walking pattern. This knowledge can, for example, serve as a reference for the imitation of natural motor control strategies in intelligent prostheses applied to walking on different terrain.

2. Methods

2.1. Instrumented staircase design

A staircase was developed that allowed the collection of kinetic data for multiple steps at different staircase inclinations [15]. It was composed of four steps and a platform at the upper end that was adjustable in height (Fig. 1). The lower three steps were instrumented with six strain-gauge force transducers each. The three components of the ground reaction force, the vertical component of the ground reaction moment and the location of the centre of pressure (COP) on the step surface were computed from the force transducer signals by static equilibrium equations [15]. The specific bearing-concept chosen in this design yielded remarkably good static properties compared to other instrumented force-platforms [16,17] used for stair walking analyses (COP error < 1.5 mm, deviation from linearity $< \pm 0.2\%$, error in accuracy $< 0.7\%$, cross talk $< 0.7\%$, see Ref. [15]). The staircase was characterised by a robust design so that dynamic properties were also satisfactory for this study. The inclination of the staircase could be readily adjusted

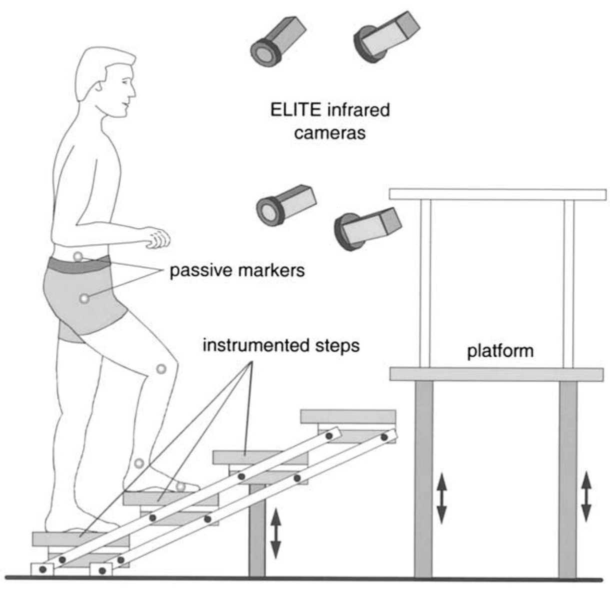

Fig. 1. Schematic drawing of staircase with platform and set-up for optoelectronic recordings. Height of staircase and platform can be readily adjusted so that the inclination can be varied between 24.0° and 42.0°. One pair of cameras was used for each body side leading to a total of four cameras.

between 24.0° and 42.0°. At an inclination of $\approx 30°$, the step dimensions were 17.0 cm (riser) by 29.0 cm (tread), which is in agreement with values proposed for the design of stairs in public environments [18]. At minimum inclination, the step dimensions were 13.8×31.0 cm and at maximum inclination they were 22.5×25.0 cm. Thus, tread size changes were small compared to changes of riser. No hand railings were necessary because only normal subjects were investigated who showed no risk of falling.

2.2. Protocol and subjects

Fig. 1 shows the experimental set-up schematically. A camera-based movement analyser (ELITE, BTS Milan, Italy; see Ref. [19]) recorded the spatial positions of 15 mm hemispherical retro-reflective markers attached to both legs at the foot (fifth metatarsal head), ankle (lateral malleolus), knee (lateral femoral condyle), hip (greater trochanter) and pelvis (upper iliac crest). A four-camera system was used to allow the kinematic measurement of both legs simultaneously. Kinematic and ground reaction data were recorded with a sampling frequency of 100 Hz.

Ten healthy male subjects of similar body height (1.79 ± 0.05 m) and weight (82.2 ± 8.5 kg), ranging in age from 24 to 34 years (28.8 ± 2.9 years) participated in the measurements. All subjects gave their informed consent for the study. All of them were free of any musculo-skeletal or neurological dysfunction. The subjects were asked to ascend and descend the stairs at minimum (24°), normal (30°) and maximum (42°) inclination. For each subject and for each inclination, ascending and descending movements were recorded for five repetitive trails.

The subjects walked barefoot at normal, comfortable speed. Stair ascent was initiated in front of the staircase on ground level, whereas stair descent started on the platform. Prior to data acquisition, the subjects ascended and descended the stairs several times until they were accustomed to the motion. Since our study concentrates on steady state walking conditions, only strides in the middle of the staircase were considered in our analysis. This means that during ascent, a stride cycle was defined starting with foot contact on the second step and ending at the next foot contact on the fourth step. During descent, the selected strides started with foot contact on the third step and ended with foot contact on the first step. Foot contact always occurred with the same foot among all subjects.

For comparison with level walking, data from 26 healthy male subjects were taken out of the data bank of the Centro di Bioingegneria gait laboratory, which were elaborated by a similar protocol as the present one [20]. Only subjects whose body size and age roughly corresponded to those of the subjects in this study (height: 1.80 ± 0.06 m, body weight: 76.7 ± 9.4 kg, age: 27.2 ± 2.6 years) were selected.

Table 1
Averaged temporal gait cycle parameters

Inclination	Cycle duration ± S.D. (s)	Stance phase ± S.D. (%)	Double support phase ± S.D. (%)
Ascent			
Minimum	1.40 ± 0.10	62.7 ± 1.8	12.7 ± 1.8
Normal	1.41 ± 0.11	63.6 ± 1.9	13.6 ± 1.9
Maximum	1.47 ± 0.11	63.7 ± 2.2	13.7 ± 2.2
Descent			
Minimum	1.20 ± 0.13	62.1 ± 2.0	12.1 ± 2.0
Normal	1.19 ± 0.10	61.2 ± 2.3	11.2 ± 2.3
Maximum	1.22 ± 0.11	59.6 ± 1.9	9.6 ± 1.9
Level walking	1.11 ± 0.05	61.1 ± 1.7	11.1 ± 1.7

2.3. Data processing

The ELITE software computed the three-dimensional Cartesian coordinates of each marker. The marker position data were low-pass filtered by a model-based bandwidth-selection procedure [21]. Internal anatomical landmarks and joint centres were estimated from the external marker positions by means of an optimisation procedure in which anthropometric measurements, joint kinematics and skeletal morphology were taken into account [22].

Hip and knee flexion–extension angles were computed in a sagittal projection. Ankle joint dorsiflexion–plantarflexion angles were computed in the plane which includes the knee, the ankle and the mid forefoot landmarks. In this way, the misleading effect of internal–external foot rotation was avoided. Angular velocities and accelerations were obtained as first and second time derivatives of the joint angle data.

Equations of motion were derived on the basis of a Newton–Euler formulation, where the lower extremity was represented including foot, shank, thigh and pelvis. Anthropometric parameters, such as mass, centre of gravity and moments of inertia of thigh, shank and foot, were estimated by regression equations provided by Zatsiorsky and Seluyanov [23]. Segment lengths were directly measured on the subjects. Joint angles, velocities and accelerations, as well as the ground reactions provided by the staircase, were fed into the equations to compute the internal joint moments, which are generated by muscles and passive structures. Mechanical power at each joint was obtained by the product of joint moment and angular velocity [24].

3. Results

3.1. Gait cycle parameters

The stance phase was between 59.6 and 63.7% of the stride duration (Table 1). During descent, stance duration percentage progressively decreased with increasing inclinations, while during ascent, the stance duration changed (increased) only slightly with stair inclinations. Differences were also observed in the stride cycle durations. They were significantly longer during ascent (1.40–1.47 s) than descent (1.19–1.22 s). They exhibited a tendency to increase with increasing inclination only during ascent.

3.2. Kinematics

Intra and inter-subject variability of the kinematic data was studied. It was observed that the patterns were reproducible and represented a 'normal' behaviour. The mean S.D. over the ten subjects (averaged over cycle time) was below 9° for the hip and below 7° for the knee and ankle joint angle. The higher variability in the hip joint is in agreement with observations from McFadyen and Winter [2]. Intra-subject variability over the repetitive trails was, in most cases, lower than the inter-subject variability.

All subjects contacted the step with the forefoot, in both climbing directions and in all inclinations (Table 2). During descent, foot orientation at initial contact was distinctly related to stair inclination and varied between −13.6° at minimum and −20.5° at maximum inclination. In contrast, during ascent, foot orientation with respect to the horizontal was about −4° and remained independent from stair inclination. On the other hand, toe clearance during ascent—measured as the maximum distance between the foot sole and the step surface—varied with stair inclination: it was lowest during normal inclination (6.5 ± 1.3 cm) and higher at minimum and maximum inclinations (7.3 ± 1.3 and 7.3 ± 1.9 cm, respectively).

Considerable differences were also observed when comparing joint angles during stair ascent and descent (Fig. 2), which is in agreement with previous studies [2]. At foot contact (0% cycle time) of stair ascent the hip and knee joints were flexed and the ankle was dorsiflexed. In contrast, at foot contact of descent, the hip was only slightly flexed, the knee was almost fully extended and the ankle joint was plantarflexed. In the subsequent phase, during ascent the hip and knee joint extended and the ankle joint globally plantarflexed, while during descent, the hip and the knee joint flexed and achieved the higher degree of flexion at late stance/early swing. The ankle joint remained dorsiflexed for most of the stance phase during descent and started to plantarflex at late stance phase. During swing phase, the maximum knee and hip flexion angles occurred later during ascent than during descent.

The joint ranges and maximum flexion angles increased with increasing inclination of the staircase. At maximum inclination, the maximum hip joint flexion during ascent was 12.4% greater than at minimum inclination. This increase was 15.7% during descent. The maximum knee joint flexion changed by 12.1% during ascent and by 14.3% during descent, the maximal ankle plantarflexion by 25.0% during ascent and by 17.3% during descent. These changes are moderate compared to the change of the staircase inclination (i.e. 75% from minimum to maximum inclination angle).

The kinematics of stair walking could be clearly distinguished from the kinematics of level walking. For example, the angular ranges were generally larger during stair walking than during level walking. Only the hip joint exhibited a range of motion that was considerably smaller during descent than during level walking (Fig. 2). The clear difference between stair and level walking is also expressed in the different foot placement (Table 2).

3.3. Ground reactions

Ground reaction force pattern during ascent and descent preserved most of the features observed during level walking (Fig. 3). However, the magnitudes of the posterior/anterior ground reaction forces were considerably smaller during stair walking. The forces produced at the beginning of the stance phase (loading response phase) were significantly higher during descent than during ascent. The influence of staircase inclination was relatively small. A significant dependency was observed only in the vertical component during early stance of descent, where the increase of reaction force between minimum and maximum inclination was 14.8%.

The COP path during both ascent and descent was limited to ≈ 10 cm in the metatarsal area. The trajectory was typically characterised by an early backward progression (relative to the foot) followed by a forward progression. Interestingly, the length of the COP path did not change significantly, although tread decreased with increasing inclination.

Table 2
Foot placement

Inclination	Mean foot orientation ± S.D.
Ascent	
Minimum	$-3.9° \pm 5.6°$
Normal	$-4.7° \pm 6.4°$
Maximum	$-4.6° \pm 6.5°$
Descent	
Minimum	$-13.6° \pm 6.1°$
Normal	$-16.6° \pm 4.7°$
Maximum	$-20.5° \pm 5.8°$
Level walking	$19.0° \pm 4.4°$

Foot orientation angle with respect to the horizontal in the sagittal plane at initial foot contact; 0° corresponds to horizontal foot orientation obtained during standing; negative values indicate foot pointing downward.

3.4. Joint moments

There was no significant dependency on staircase inclination when comparing the joint moments during the swing phase (Fig. 4). In contrast, during the stance phase considerable differences were observed in the hip and knee joint moments. The hip joint produced a flexion moment during descent, whereas during ascent, an extension moment was produced during most of the stance phase. Despite the different amounts of knee flexion between the ascending and descending movement (Fig. 2), the knee joint moments were very similar for ascent and descent at the beginning of the stance phase (0–30% cycle time). However, they dramatically differed in the second half of the stance phase (35–60% cycle time). During stair descent there was a second peak that further tended to extend the knee, while during ascent the knee moments became relatively small and reversed in sign. Ankle joint moments were similar during ascent and descent, except that the first peak was higher during descent and the second peak was higher during ascent.

The maximum moment values increased with increasing stair inclination at the knee during both ascent (10.6%) and descent (18.4%) and at the hip during ascent (37.9%). During descent, the maximum hip joint moment became even smaller with increasing inclination (-27.1%). At the ankle mainly at the beginning of the stance phase (first peak), a significant dependency of the joint moment on the inclination could be observed. During this phase, the ankle moment increased with increasing inclination for both ascent (12.8%) and descent (18.7%).

The joint moment patterns and ranges of stair walking were relatively different to those of level walking.

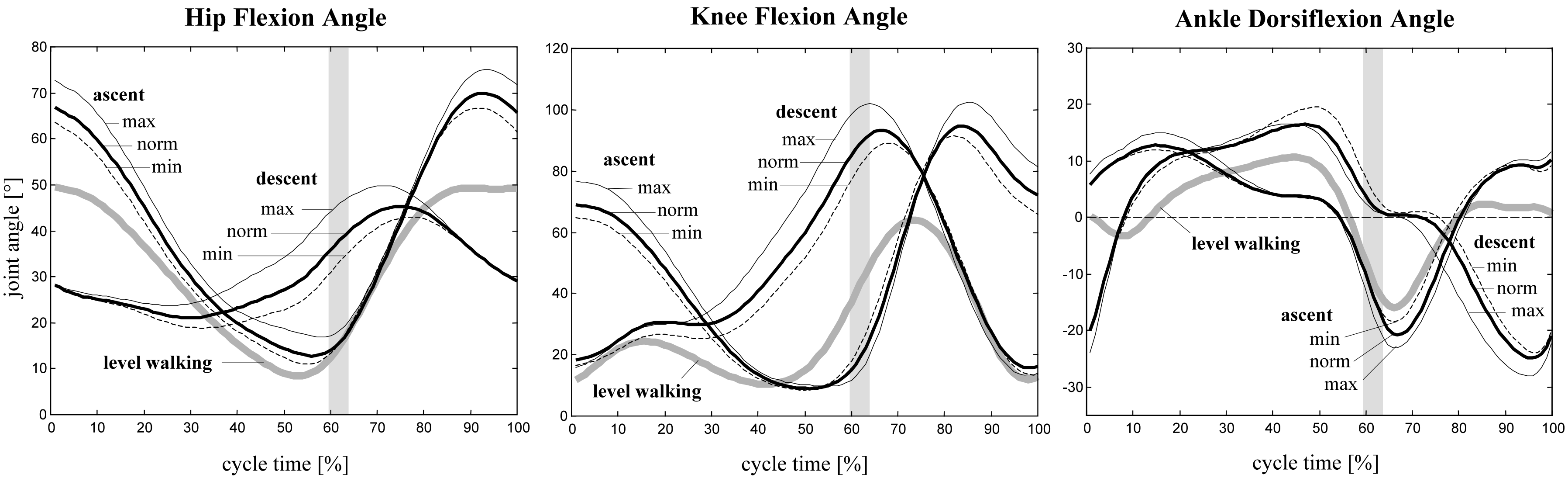

Fig. 2. Joint angles during ascent and descent at minimum, normal and maximum inclinations and during level walking averaged over all subjects. The cycle starts with foot contact. The vertical grey bar indicates toe off, thus splitting the entire stride into stance and swing phase. The width of the bar corresponds to the variations of stance phase durations (59.6–63.7%, see Table 1) observed for the different inclinations at descent and ascent.

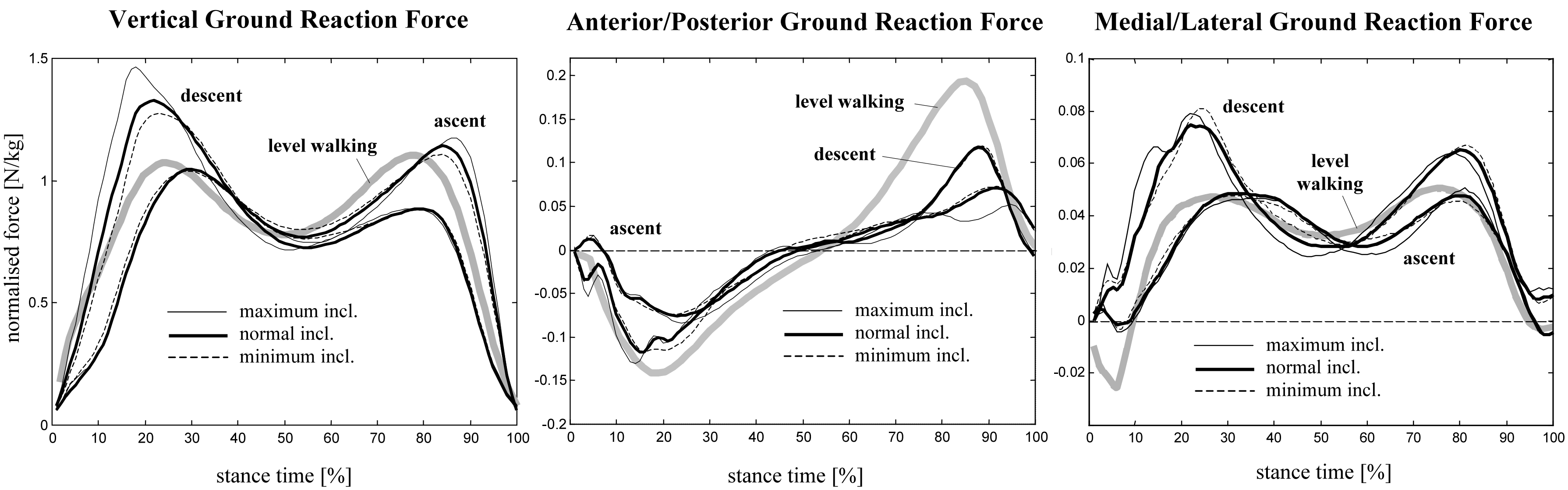

Fig. 3. Ground reaction forces during ascent and descent at minimum, normal and maximum inclinations and during level walking averaged over all subjects. Forces are normalised by body weight. The cycle starts with foot contact.

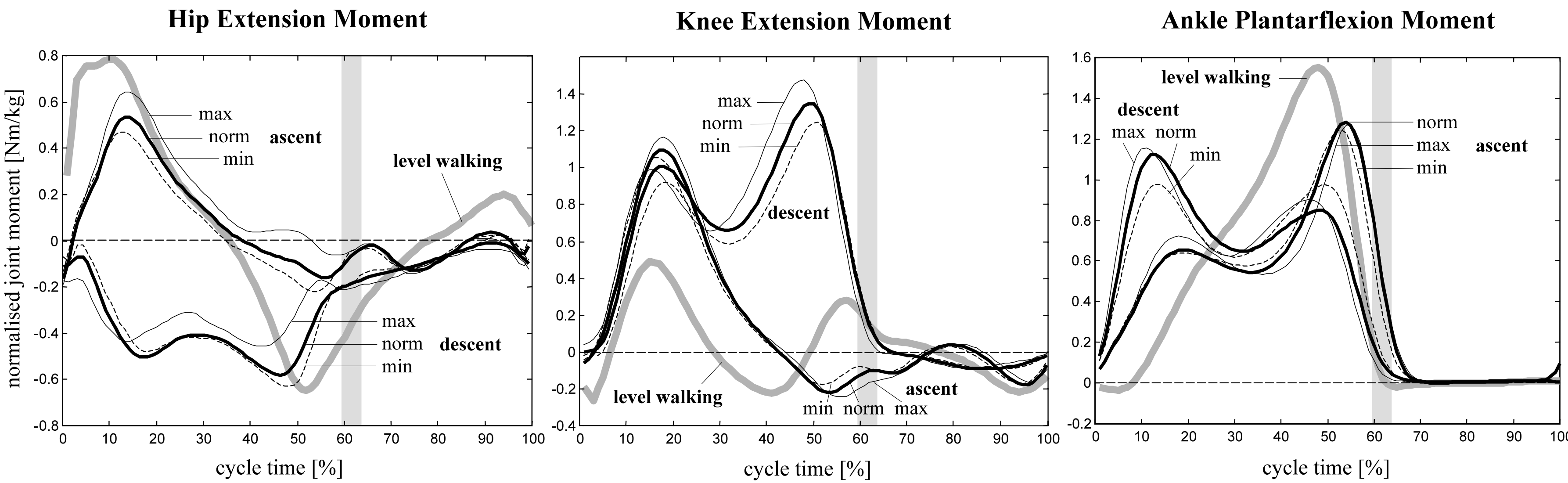

Fig. 4. Joint moments during ascent and descent at minimum, normal and maximum inclinations and during level walking averaged over all subjects. Moments are normalised by body weight. The cycle starts with foot contact. The vertical bar indicates toe off for the different inclinations (see legend to Fig. 2).

In particular, the maximum knee extension moment was up to three times greater than the knee moment generated during level walking.

3.5. Joint powers

During ascent, all the joints produced energy (positive power) during most of the stride phase (Fig. 5). The knee and hip joint powers reached their maximum at the beginning of the stance phase at $\approx 14–20\%$ cycle time. In the hip, a second lower peak was observed during the swing phase. The ankle joint exhibited maximum power production at the end of the stance phase (53–59% cycle time), not only during ascent but also descent.

During descent, the joint powers were predominantly negative, i.e. energy was absorbed. Only the hip joint showed a remarkable phase of energy production, with a peak of power at $\approx 40–55\%$ cycle time. This maximum value was, however, roughly only 30% of the value observed during ascent. The maximum (negative value) of the knee joint power occurred at $\approx 48–54\%$ cycle time and was—in absolute value—higher than the knee power generated during ascent. The ankle joint absorbed power during descent at the beginning of the stance phase (8–10% cycle time).

Joint powers were more dependent on staircase inclination than joint angles and moments. Absolute joint power maximums increased with increasing staircase inclination. The strongest influence was observed at the hip joint during stair ascent (51.7% increase of maximum joint power from minimum to maximum staircase inclination) and in the ankle joint during descent (67.3%) and ascent (45.4%). Smaller values were observed in the knee joint during ascent (25.1%) and descent (26.3%) and the hip during descent (24.3%). In contrast, the time when maximum joint power occurred did not depend appreciably on the inclination.

Only some similarities were observed between stair and level walking power patterns, for example, at the ankle during late stance and at the hip during early stance. Distinct differences were observed in the knee. Neither shape nor magnitudes corresponded well to the power patterns obtained from level walking. During descent, the maximum power absorption at the knee was up to 3.8 times greater than during level walking.

4. Discussion

4.1. Differences between ascent and descent

A fundamental consideration, already pointed out in the work of McFadyen and Winter [2], is that the ascending task consists primarily of a transfer of muscle energy into potential (gravitational) energy of the body, whereas during descent, the potential energy has to be dissipated (absorbed) by the muscles. During descent, this process is achieved first by a transfer of potential energy into kinetic energy. This takes place during the swing phase, which is a rather ballistic movement, since joint moments and powers remain small (see Figs. 4 and 5). Almost all the kinetic energy accumulated has to be absorbed at the subsequent foot contact. A direct consequence is that during the loading response phase (also called 'pull up' phase [2]) the ground reaction forces are considerably higher during descent than during ascent (Fig. 3). In contrast, during ascent the main phase of energy production takes place at the 'push-up' phase, where the ground reaction forces are higher than during descent.

These effects of potential energy production and dissipation are also expressed by the phases of positive and negative joint powers observed during ascent and descent, respectively (Fig. 5). During ascent, a large power production in hip and knee joint occurred right after foot contact and peaked at $\approx 14–20\%$ of the stride cycle time (pull up phase). Power production in the ankle appeared considerably later, with a peak at $\approx 58\%$ of the stride cycle (push up phase). Interestingly, during this phase there was no significant activity (neither moment nor power) in the hip and knee joint, unlike that observed in EMG recordings [14]. If we consider that 50% of the ipsilateral stride cycle corresponds to the initial contact of the contralateral side, it can be seen that push up at the ankle joint occurs prior to pull up at the contralateral knee and hip joints. Despite this temporal separation it seems that the two phases—push up and pull up—are a part of the same mechanism of body lifting: looking at the joint angles of the stance leg, the hip and knee joints produce joint powers that lead to maximum extension at the end of the stance phase (40–60% of stride cycle), while the foot is resting on the ground (foot orientation remains $\approx 0°$). At that time, knee and hip cannot raise the body any more. As soon as the contralateral foot has approached the next step, the ipsilateral ankle plantarflexes and produces the push up power that supports the transfer of the body weight to the leading limb and reduces the need for higher hip and knee joint moments.

Stair descent shows a strong ankle power absorption just after foot contact (at 8–10% of stride cycle, accentuated by a moderate knee power absorption at 10–13%), when the knee and ankle joints are being extended. However, this phase alone cannot be efficient as 'shock absorbers', where the entire potential energy is dissipated. Rather, it appears that the three joints of both sides are activated in a sequence aimed at sharing energy absorption among them. Prior to the ankle

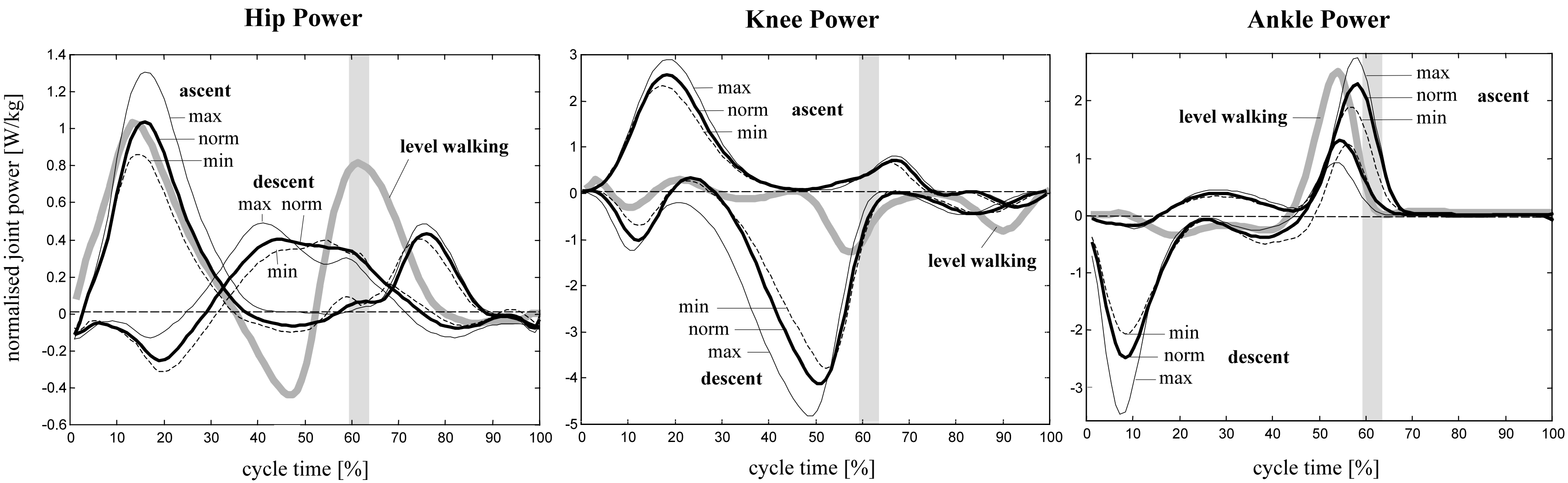

Fig. 5. Joint powers during ascent and descent at minimum, normal and maximum inclinations and during level walking averaged over all subjects. Powers are normalised by body weight. Positive powers indicate concentric, negative powers eccentric phases. The cycle starts with foot contact. The vertical bar indicates toe off for the different inclinations (see legend to Fig. 2).

joint, the contralateral knee joint already starts to absorb a significant amount of energy at about -2 to 4% stride cycle (again, assuming that 50% of the ipsilateral stride cycle corresponds to the initial contact of the contralateral side). Following the ankle joint, the hip and then the knee show peak power absorptions at 16–22% and 50–54% stride cycle, respectively. Finally, the contralateral side finishes the gait cycle with energy absorption at the ankle (58–60%) and hip (66–72%). Note that during these phases, the antigravity muscles perform eccentric contractions, i.e. they generate force while they are lengthened. These are situations that have been demonstrated to be of potential risk of muscle fibre damage when speed or loads are becoming too high [25].

The interpretation of the different joint moment signs during ascent and descent (Fig. 4) must be related to the direction of the line of action of the ground reaction force, which is responsible for the moments generated in the joints. In the first half of the stance phase, this line is slightly backward oriented (negative horizontal component, see Fig. 3) and becomes forward oriented in the second half of the stance phase. Since the COP (where, by definition, the ground reaction line of action traverses the ground) is located in the metatarsal area (see Section 3.3), the ankle joint produces a plantarflexion moment to counteract the ground reaction during the entire stance phase. The large knee flexion that occurs during late stance of descent (Fig. 2) causes the centre of the knee joint to move forward. Thus, the ground reaction force tends to flex the knee which is responded by an increase of knee extension moment during this phase of descent (Fig. 4). In contrast, the knee extension that occurs at late stance of ascent yields a reduction of the distance between the ground reaction line of action and the knee joint centre, so that the extension moment produced by the knee joint is decreasing and changing in sign in the second half of the stance phase. A similar argumentation can explain the moment-time-course at the hip joint during ascent and descent (Fig. 4), where differences among the two conditions are even more distinct.

4.2. Influence of staircase inclination

Our kinematic and kinetic data obtained at normal inclination correspond well with patterns described in the literature [1–5]. The outstanding contribution of this study is the presentation of data that depict the influence of different inclinations. The stair inclination angles investigated in this study reflect a typical range of staircases that we encounter in daily life. In most public environments, the step dimensions are very similar to those used at medium inclination in this study.

Although there was a significant dependency of most gait parameters on staircase inclination, the intensity of this dependency was different. For example, there was only a little to moderate influence of the inclination angle on joint angle patterns (Fig. 2), gait phase parameters (Table 1) and joint moment patterns (Fig. 4). Angular ranges of all joints increased with increasing inclination angles. This is consistent with the need for a higher elevation of the foot at increased step heights and agrees with some basic findings presented in [13].

The largest differences were observed in the joint power patterns. Maximum joint powers in the hip and ankle changed with inclination up to $\approx 67\%$. This change can be related to the varying amount of potential energy that has to be produced (during ascent) or absorbed (during descent) by the muscles in order to surmount stairs with different inclinations. This finding is consistent with observations made with the ground reaction force (Fig. 3) as well as in EMG recordings [14].

To assess the influence of inclination also quantitatively, some characteristic gait values were chosen and plotted versus increasing inclination angles (Fig. 6). Least squares were used to fit a regression to each set of three characteristic values of descent (at inclinations $-42°$, $-30°$, $-24°$) and ascent (at inclinations 24°, 30°, 42°). Linear regressions have shown to approximate the inclination dependent tendencies remarkably well. By looking at the slopes of the regression lines, the above-mentioned qualitative findings can be confirmed that most gait parameters increase with increasing stair inclination angle. It can be further seen that joint powers and foot placement during descent showed a very strong dependency on stair inclination.

The increased toe clearance (Section 3.2) and cycle durations (Table 1) at minimum and maximum inclination angles could be interpreted as signs that stair climbing at extreme inclinations is a particular challenge for the human motor control system compared to climbing a normal stairs. Further investigations with a higher range of inclination angles and additional recordings of muscle activity could shed light into this hypothesis.

4.3. Comparison with level walking

One can assume that ascending and descending motions are to be considered as particular evolution of level walking, i.e. the kinematic and kinetic patterns of level walking are related to the patterns of ascent and descent in a particular manner. However, only little signs could be found that indicate an adaptation or shift in the motor patterns when moving from level to stair walking. For example, the vertical ground reaction force during level walking gained values between those of ascent and descent. Furthermore, stair ascent and level walking showed some agreement concerning the

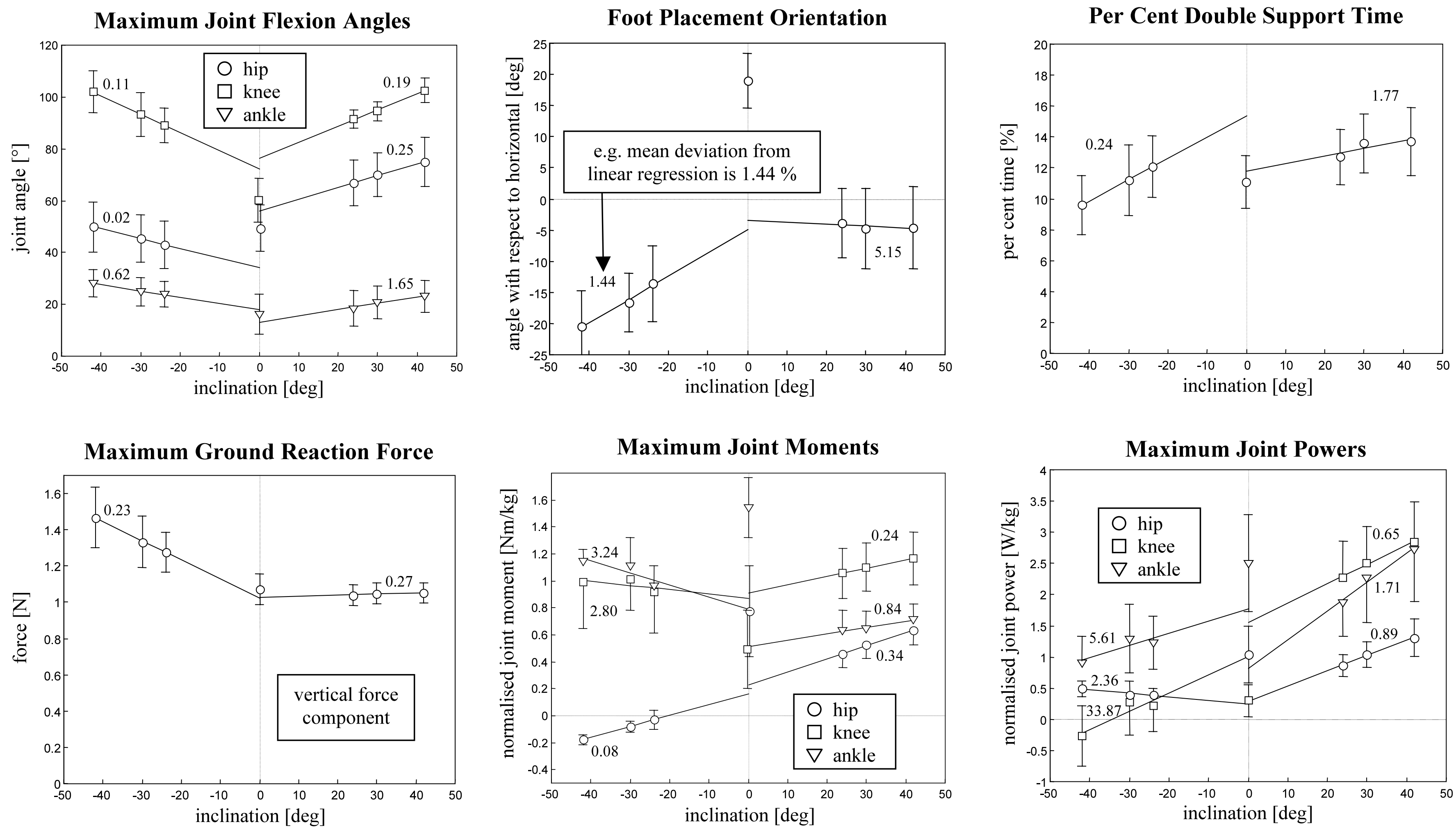

Fig. 6. Characteristic kinematic and kinetic group averaged values (circles, squares, triangles) and their S.D. versus stair inclination angle. Descending stairs was represented by negative inclination angles and level walking by 0°. Note that the hip and knee moment values are given by the first local maximum values taken from the time patterns (Fig. 4). Linear regressions are depicted by solid lines. The numbers indicate the mean deviation from the linear regression in percent.

energy producing phases during the loading response phase of the hip and the push up phase of the ankle (Fig. 5). No significant correlation was observed between the joint powers during descent and level walking: the typical sharing of energy absorption among the leg joints was not observed during level walking.

In general, the gait patterns (Figs. 2–5) did not change in a progressive way, when comparing descent at decreasing inclinations ($-42°$, $-30°$, $-24°$), level walking and ascent at increasing inclinations (24°, 30°, 42°); differences in the gait data between minimum and maximum inclination were in general considerably smaller than the differences between level walking and ascent or level walking and descent. This can be confirmed also when considering certain characteristic values, such as maximum angles, moments or powers (Fig. 6). In most cases, the intersection of the extrapolated linear regression with the vertical line at 0° inclination clearly deviated from the characteristic values observed at level walking. A remarkably high deviation was observed for the foot placement; only during level walking was heel contact observed.

During ascent, the different foot placement can be explained by the fact that only a forefoot contact keeps the ankle in a natural angular range. Due to the strong knee flexion during ascent, which is necessary to lift the foot onto the next step (Fig. 2), heel contact would require a relatively high dorsiflexion angle. However, then an increased dorsiflexor force would be necessary to compensate for the high passive elastic plantarflexion moment observed in this range [26]. During descent, the ankle plays an important role in the sharing of energy absorption (Section 4.1). Thus, a possible reason for the forefoot contact during descent is that only in this ankle posture a downward rotation of the foot is possible, during which a considerable amount of energy can be absorbed.

A consequence of the forefoot placement during stair climbing compared to the heel placement during level walking is that the ankle has to produce a higher plantarflexion moment during the initial stance phase (0–20% cycle time, Fig. 4) in order to keep the heel lifted. This finding agrees with the observation that the COP path starts at the anterior part of the foot (see Section 3.3).

The above-mentioned findings lead to the assumption that there is a certain inclination angle or angular range, where the subjects switch their gait patterns and, thus, their motor control strategy between level and stair walking. This change must occur at an inclination angle below 24° and might be related to the condition at which initial foot placement switches from heel contact to forefoot contact. Further studies are necessary to confirm this hypothesis and detect the inclination at which this switch between gait patterns takes place.

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