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Influence of body mass and lower limb length on knee flexion angle during walking in humans

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Abstract. Despite abundant knowledge about the relationship between body size (i.e., body mass, lower limb length) and limb posture during locomotion on the level of interspecies variability, little is known about variation on the intraspecific level. We used an experimental approach to evaluate the relationship between body size and knee posture during walking in humans at specific gait events and at each percentage point of normalized stance phase. We detected significant negative correlation between knee flexion angle and body mass at the second peak of the vertical ground reaction force, but, in contrast to a previous study, we found no significant relationship between knee flexion angle and lower limb length. Although not significant, strengthened correlations between knee flexion angle and lower limb length were detected at late stance phase and these coincide well with the strengthened correlations between knee flexion angle and body mass. Our findings support the view that body size influences limb posture during locomotion even on the intraspecific level. In humans, larger individuals tend to use more extended knee postures in late stance of walking than do smaller individuals.

Key words: size, posture, locomotion

Introduction

It is widely accepted that body size (i.e., body mass, lower limb length) influences limb posture during locomotion on the level of interspecies variability (Biewener 1989a, b, Gatesy & Biewener 1991, Reilly et al. 2007). However, little is known about the relationship between body size and limb posture during locomotion on the intraspecific level. Only a few studies have addressed this problem, among which Gruss (2007) was particularly interested in lower limb length and the knee flexion angle during walking in humans.

Limb posture, defined as the relative position of the limb segments, has a major effect on the cost of locomotion and the forces acting in the limb bones and muscles through its influence on the magnitude of the bending moments exerted about the joints and bones (Gray 1968, Biewener 1983, 1989a, Reilly et al. 2007). Changes in limb posture influence the relative position of the ground reaction force (GRF) vector and the length of its moment arm (R). Through its influence on R, limb posture influences the external moments exerted by GRF about the point of interest along the limb (joints, bones), since external moment = GRF × R (Fig. 1). The greater the external moment, the greater must be the internal moment generated by muscles to counteract the external moment in order to keep or change the desirable limb posture. Since magnitude of the GRF increases in proportion to body mass and R increases in proportion to lower limb length, animals of increasing body size adjust their limb postures in order to keep the external moments on the level that their muscles and bones can withstand (Biewener 1983, 1989a, b). Such size-dependent changes in posture have been detected not only on the level of interspecific variation, but also

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among closely related species and to a limited extend also on the intraspecific level.

Polk (2002) investigated the effect of body size on limb posture and joint moments during walking in monkeys from three closely related species. He concluded that individuals with greater body mass keep their knees and elbows more extended and have lower joint moments at mid-stance. Comparison between individuals with approximately the same body mass but different limb length showed that limb length has similar influence on limb posture and joint moments as does body mass: individuals with longer limbs kept their knees and elbows more extended and they had lower joint moments than did those with shorter lower limbs.

The influence of body size on limb posture during locomotion has been investigated also in humans. Some insight as to the influence of body mass on limb postures can be taken from studies comparing lean and obese samples of individuals, and further there is the study by Gruss (2007) investigating the influence of lower limb length on knee flexion angle. DeVita & Hortobágyi (2003) studied the kinematics of lean and obese samples of human individuals and concluded that obese individuals had less knee flexion throughout the early stance (first half of stance phase) than did lean individuals; the difference was about 8°. Furthermore, knee internal moments were surprisingly similar or even less in obese participants compared to those of lean participants. This suggests that keeping the knee more extended can be an effective strategy for moderating knee extensor moments. Similar results were obtained when comparing lean vs. obese children (Gushue et al. 2005). On the other hand, some studies did not detect differences in knee flexion angle between lean and obese humans and reported significantly higher knee internal moments in obese participants than in lean participants (Browning & Kram 2007, Shultz et al. 2009). Thus, we could summarize that humans with high body mass and high body mass index (BMI) are able to moderate knee moments during early stance of walking by using more extended knee postures, but it is not clear when this moderating mechanism is used.

Gruss (2007) investigated the influence of lower limb length on limb posture at the knee in a sample of non-obese humans. She found that in late stance (second half of the stance phase), and particularly at the second peak of vertical ground reaction force and at the peak of propulsive ground reaction force, the lower limb length was negatively correlated with knee flexion angle. In other words, individuals with longer lower limbs kept their knees more extended during the late stance. Unfortunately, Gruss (2007) did not report the influence of body mass on kinematics and so it is not clear if body mass has some influence independent of lower limb length in early stance as could be expected based on the results of studies comparing lean vs. obese samples. On the other hand, a claim has been raised that a detected relationship between lower limb length and knee kinematics in late stance could be confounded by covariation with body mass (Shaw & Stock 2011). So, there exist at least two reasons to repeat the experiment carried out by Gruss (2007) with the inclusion of another variable besides lower limb length, that variable being body mass.

In this study, we pursue two objectives. First, we aim to verify the relationship between lower limb length and knee flexion angle during the late stance phase

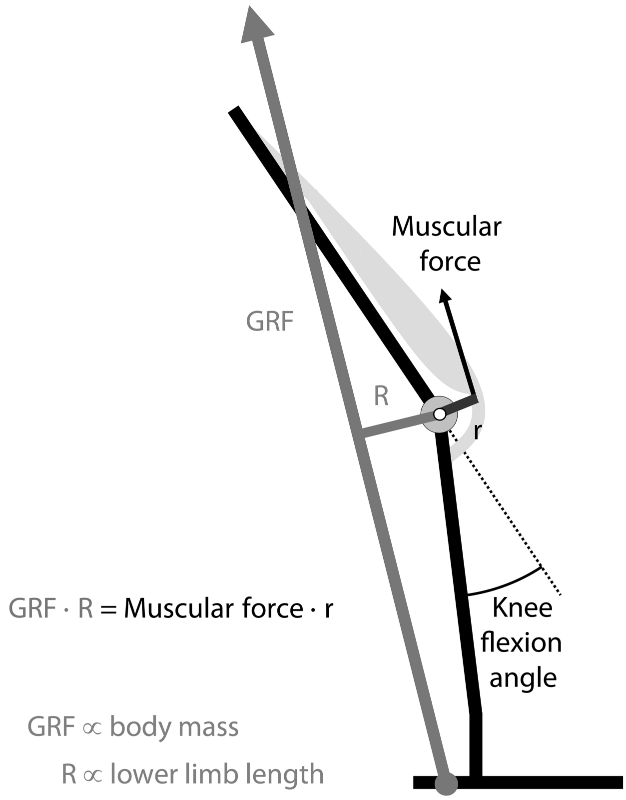

Fig. 1. *Simplified diagram of the forces acting at the knee in the sagittal plane. To keep the knee at a given angle, muscular force must counterbalance the ground reaction force (GRF) through a system of levers (R, r). Since GRF scales approximately with body mass and R scales with lower limb length, larger and longer-limbed individuals need more muscular force to counterbalance the GRF unless they 1) elongate the r; and/or 2) shorten the R by reducing knee flexion.*

of walking as detected by Gruss (2007) in another sample of humans while examining it in more detail – not only at a limited number of gait events but also at each percentage point of normalized stance phase. Second, we aim to explore the relationship between knee flexion angle and body mass with the expectation to find negative correlation between these two variables in early stance.

Material and Methods

Twenty-six adults (12 females, 14 males) with no history of lower limb injuries participated in this study. Subjects were non-obese (body mass index under 30 kg/m^2) and between 19 and 38 years of age. All subjects signed a consent form approved by the Institutional Review Board of Charles University, Faculty of Science prior to examination.

Gait and anthropometric data for each participant were collected in the biomechanical laboratory of CASRI – Sports Research Institute of the Czech Armed Forces during one single session per participant. Each session consisted of three parts: 1) the participant's familiarization with a treadmill, 2) anthropometric data collection, and 3) gait data collection.

During the first part of the session, participants walked and ran on the motorized treadmill at variable speeds for approximately 25 minutes to become accustomed to the treadmill. We identified each participant's most comfortable walking speed (preferred speed) during this part of the session.

The second part of the session lasted 30 to 45 minutes, during which we measured a set of the participant's body dimensions and the participant had time to rest before the third part involving gait data collection. From the set of measured body dimensions, we used in this study stature, body mass and length of dominant lower limb. Stature and lower limb length were measured by anthropometer in millimetres. Body mass was measured using a digital weighing scale in kilograms. Lower limb length was measured as the sum of thigh length and shank length. We measured thigh length as the distance between the greater trochanter and lateral epicondyle. We measured shank length as the distance between the medial midpoint of the knee and the distal tip of the medial malleolus. In this study, we use the term "dominant lower limb" for that limb which is used for activities requiring fine manipulation and focused attention, generally for mobility (Peters 1988, Sadeghi et al. 2000). The dominant lower limb was identified by questionnaire inquiring as to the preferred lower limb in different activities (kicking a ball, hopping on one foot, stepping on a chair, and stamping on an object).

During the third part of the session, the actual collection of gait data, participants walked on the treadmill at their previously identified preferred speed. We used synchronized pressure measuring insoles (Pedar, Novel) to measure kinetics and an optical motion capture system using 10 infrared cameras (Qualisys) to measure kinematics. The recording frequency was set to 100 Hz for both devices. Each participant wore his or her own t-shirt and sports shorts and was provided uniform neoprene shoes (Hiko Softy) with a thin sole in order to hold the measuring insoles and still be able to imitate barefoot walking.

The modified calibrated anatomical system technique (CAST) marker set (Cappozzo et al. 1995) was used to track lower limbs kinematics. The pelvis was tracked using four markers placed on the anterior superior iliac spine (ASIS) and posterior superior iliac spine (PSIS) landmarks, thigh and shank segments were each tracked using a four-marker cluster, while foot segments were each tracked using three markers placed on the tuber calcanei and on the heads of the first and fifth metatarsals. Additionally, the location of nine bony landmarks per limb in relation to marker clusters were found by manual palpation and recorded using a digitizing pointer. A seven-segment model was built in Visual3D (C-Motion). Local anatomical frames for each segment were defined following International Society of Biomechanics recommendations (Wu et al. 2002). The hip joint centre, needed for definition of the femoral mechanical axis, was estimated using a functional approach (Schwartz & Rozumalski 2005, Begon et al. 2007).

For each participant, recordings 10 seconds long were used for further analyses. The raw data from Novel and Qualisys were imported into Visual3D software and filtered using a fourth-order low-pass Butterworth filter with a 6 Hz cut-off frequency. Four to 10 strides were analysed for each participant and were selected on the basis of good quality of the marker trajectories and ground reaction forces.

Since no standard technique for identifying the neutral position of the knee angle has been defined (Perry & Burnfield 2010), we used no adjustments for knee flexion angle in this study. Our zero knee flexion angle is achieved when the angle between the femoral mechanical axis and tibio-fibular mechanical axis is 180°. No information about whether or not the knee flexion angle was adjusted to neutral position is provided by Gruss (2005, 2007).

The knee flexion angle of the dominant lower limb was evaluated at the particular gait events and at each percentage point of the normalized stance phase.

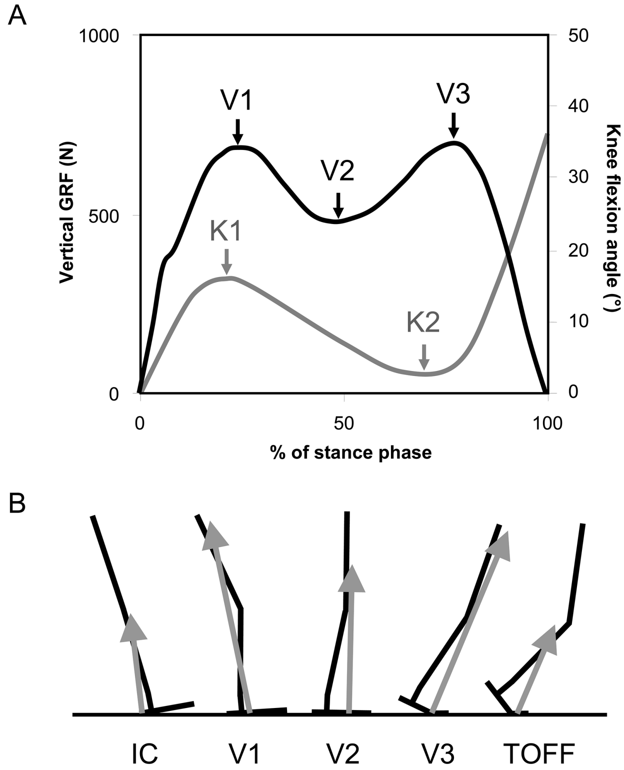

Fig. 2. *A. Kinetic events (V1, V2, V3) indicated on the trace of vertical ground reaction force (black line) and kinematic events (K1, K2) indicated on the trace of angular displacement of the knee (grey line). B. Scheme of the ground reaction force (GRF) vector and limb segments orientation at the kinetic events. IC, initial contact; V1, first peak of vertical ground reaction force; V2, mid-stance minimum of vertical ground reaction force; V3, second peak of vertical ground reaction force; TOFF, toe off; K1, maximum knee flexion angle at the first half of stance; K2, minimum knee flexion angle at the second half of stance.*

We used the following gait events in this study (Fig. 2): initial contact (IC), first peak of vertical ground reaction force (V1), mid-stance minimum of vertical ground reaction force (V2), second peak of vertical ground reaction force (V3), and toe off (TOFF); maximum knee flexion angle at the first half of stance (K1), and minimum knee flexion angle at the second half of stance (K2). We were not able to provide pressure measuring insoles to four of our female participants because of the small size of their feet and so their kinetics could not be measured. The sample size at events V1, V2 and V3 was therefore reduced to n = 22, whereas at all other gait events and at each percentage point of normalized stance phase sample size is n = 26. To avoid similar sample size reduction at initial contact and toe off, we determined these events using a velocity-based detection algorithm (Zeni, Jr. et al. 2008) verified by visual inspection. In order to be able to compare our results to results of Gruss (2007), we used pooled sample of females and males for analyses in this study.

The Mann-Whitney *U* test was used to test for differences between our sample and that of Gruss (2005, 2007), because two variables were non-normally distributed in Gruss' sample (body mass, walking speed) and one variable was non-normally distributed in our sample (body mass index). We used Pearson correlation whenever possible to evaluate association between size variables (body mass, lower limb length) and the knee flexion angle at gait events and at each percentage point of normalized stance. Further, we used Kendall's rank correlation at gait events to be able to compare our results to those of Gruss (2007). Additionally, we used linear regression analysis to visualize the selected relationships. We ran the Shapiro-Wilk test prior to other statistical analyses, which showed that all variables in our sample were normally distributed except the knee flexion angle at the event V1.

Results

Descriptive statistics for the samples used in this study are given in Table 1. Mean stature in our sample (1743

Table 1. *Descriptive statistics for the samples used in this study with significant differences between the samples indicated by asterisks (Mann-Whitney U test).*

Sample		Stature (mm)	Body mass (kg)	Body mass index (kg/m^2)	Lower limb length (mm)	Preferred speed (km/h)
Our sample	Mean	1743	68.7	22.4	823	4.72
(n = 26)	SE	22	2.9	0.5	13	0.11
Gruss (2007)	Mean	1706	72.0	24.4	760	4.93
(n = 27)	SE	22	3.1	0.6	14	0.10
Significance		ns	ns	**	**	ns

SE, standard error of the mean.
** $p < 0.05$; ** $p < 0.01$; ns, non-significant difference.*

Table 2. Mean and standard error of the mean (SE) of the knee flexion angles at the gait events of our sample and sample of Gruss (2007). (Dash indicates no data available for specific event. For definition of gait event abbreviations, see Fig. 2 caption.)

Sample		Knee flexion angle at the event (°) IC	V1	V2	V3	TOFF	K1	K2
Our sample	Mean	−2.4	14.7	6.2	4.4	38.3	14.9	1.4
(n = 26)[1]	SE	0.8	1.1	0.8	0.6	0.7	1.0	0.5
Gruss (2007)	Mean	-	27.8	13.4	12.5	-	-	-
(n = 27)	SE	-	0.8	0.7	0.8	-	-	-

[1] At events V1-3 the sample size is reduced to n = 22.

mm), although 37 mm higher, is not significantly different from mean stature for the sample of Gruss (2005, 2007). Mean body mass in our sample is 68.7 kg and although lower it is not significantly different from that of Gruss' sample. Mean body mass index in our sample (22.4 kg/m^2) is significantly different from that of Gruss' sample (Mann-Whitney *U* test, p = 0.010), the latter being 24.4 kg/m^2 and which is on the boundary of overweight. Mean lower limb length in our sample (823 mm) is significantly different from that of Gruss' sample (Mann-Whitney *U* test, p = 0.004), the mean lower limb being 63 mm longer in our sample. Mean preferred walking speed in our sample (4.72 km/h) is not significantly different from that of Gruss' sample. Our sample does not differ overall from that of Gruss (2005, 2007) in absolute body dimensions but rather in body proportions, as individuals in our sample have lower body mass relatively to stature and shorter lower limb relative to stature. It is not clear how these differences in body proportions could influence the examined relationships between knee flexion angle and body mass or lower limb length.

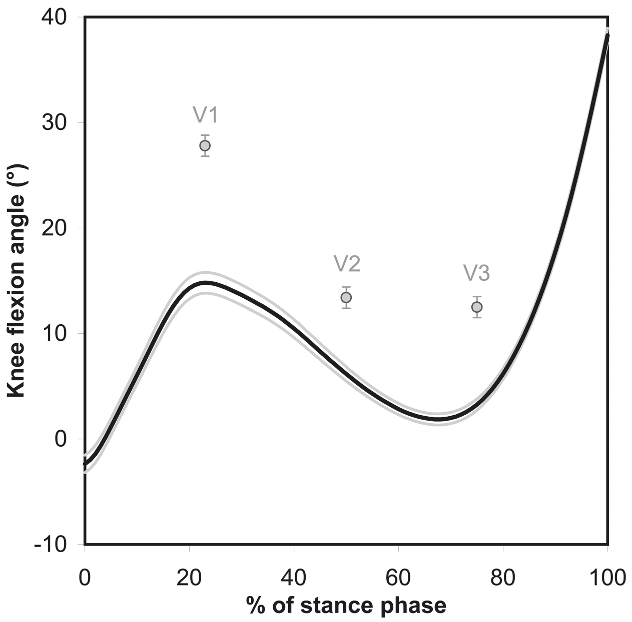

Fig. 3. Mean knee flexion angle (black line) and standard error of the mean (grey lines) of our sample during the stance phase. Mean knee flexion angle (grey circles) and standard error of the mean (error bars) of Gruss' (2007) sample at events V1, V2 and V3 is shown for comparison.

Mean knee flexion angle of our sample at the gait events and at each percentage point of normalized stance phase are given in Table 2 and Fig. 3, with values for gait events in Gruss' sample provided for comparison. In our sample, mean knee flexion angle rises from slight hyperextension at initial contact to early stance peak flexion of 14.8° at 23 % of normalized stance. After the peak of knee flexion in early stance, the knee joint extends to almost full extension (1.8°) at 68 % of normalized stance. Knee flexion then increases again to 38.3° at toe off. Gruss (2005, 2007) published mean knee flexion angle only at three gait events equivalent to our events V1, V2 and V3. At all three events where comparison is possible, knee flexion angle is considerably higher in Gruss' sample than in our sample, with the greatest difference being 13.1° at V1.

Coefficients of correlation (Pearson's r and Kendall's τ) between body mass or lower limb length and the knee flexion angle at gait events are given in Table 3. Significant negative correlation between body mass and knee flexion angle was identified at event V3 in our sample, but the relationship is rather weak (Pearson's r, r = 0.44, n = 22, p = 0.038) and was not confirmed by Kendall's test of ranked correlation (n = 22, p = 0.108). Correlation between lower limb length and knee flexion angle was not significant at any event in our sample, which is inconsistent with the results of Gruss (2007). She had found significant correlation between lower limb length and knee flexion angle at event V3.

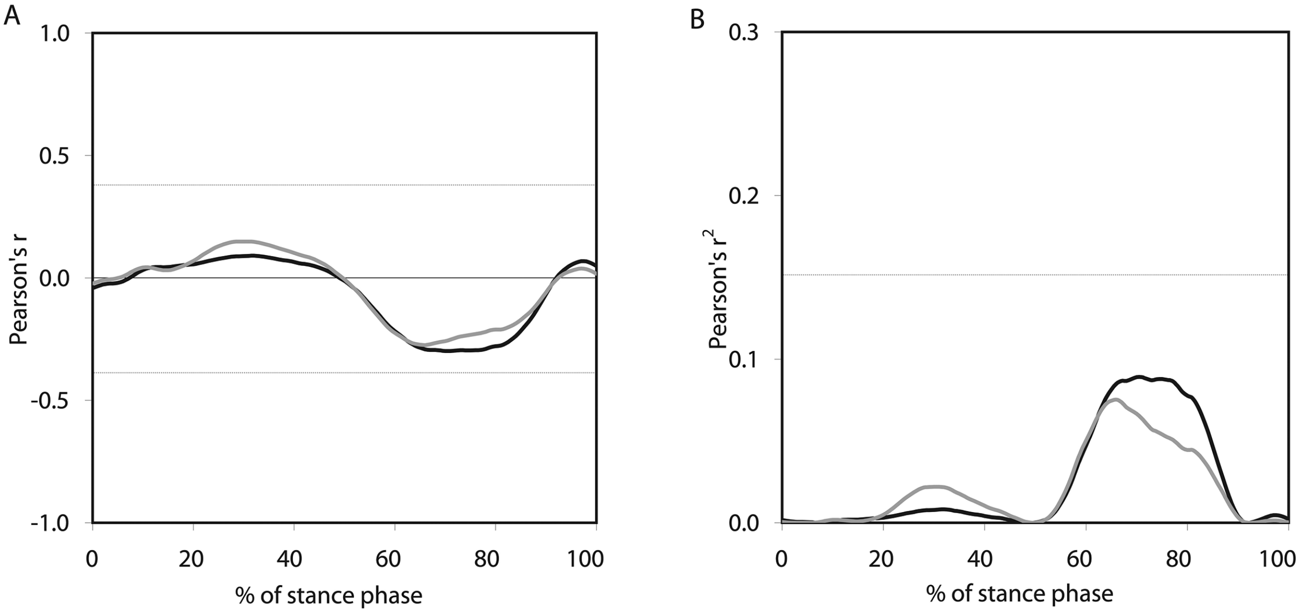

Fig. 4. *Coefficient of correlation (A) and coefficient of determination (B) between body mass (black line), lower limb length (grey line), and the knee flexion angle at each percentage point of normalized stance phase, with 0.05 p-level indicated by grey dotted lines.*

Correlation between body mass or lower limb length and the knee flexion angle at each percentage point of normalized stance phase is shown in Fig. 4. Neither body mass nor lower limb length is significantly correlated with knee flexion angle at any percentage point of normalized stance phase. In early stance, however, there is a slight tendency for positive correlation between the knee flexion angle and both body mass and lower limb length, the strongest correlation coefficients being 0.09 at 32 % of stance for body mass and 0.15 at 31 % of stance for lower limb length. During late stance there is seen to be a stronger tendency for negative correlation between the knee flexion angle and both body mass and lower limb

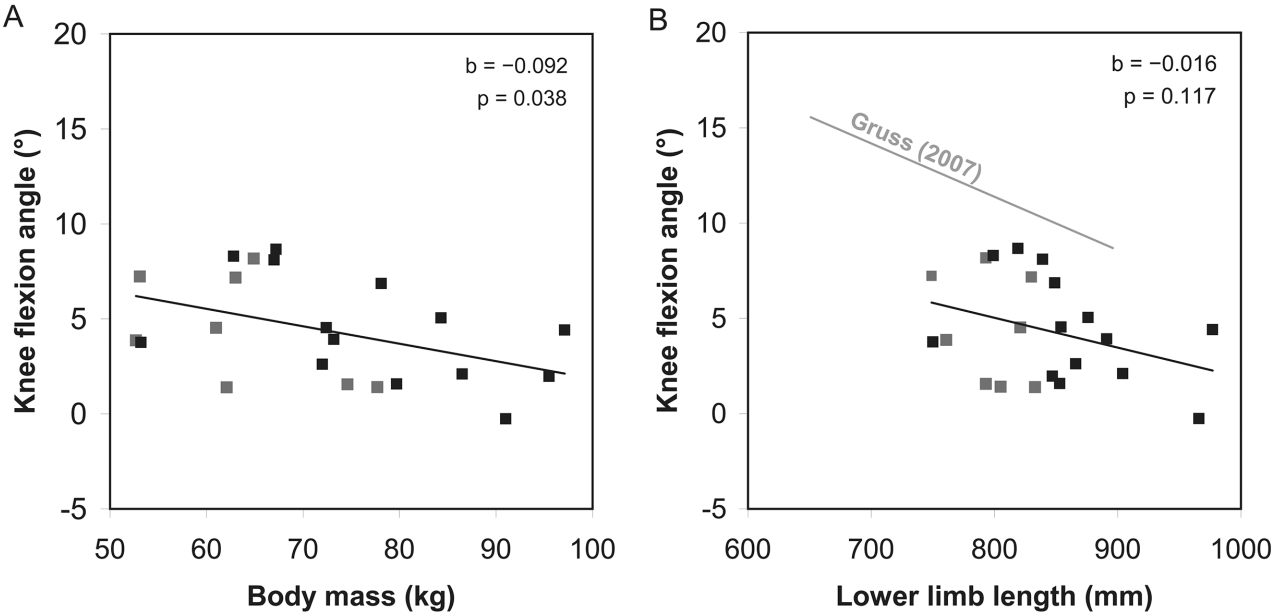

Fig. 5. *Linear regression (black line) of the knee flexion angle on body mass (A) and lower limb length (B) at event V3. Pooled sample of females (grey squares) and males (black squares) was used for regression analyses. Slope of linear regression based on data by Gruss (2005, 2007) is shown for comparison (grey line).*

Table 3. *Kendall's Tau (A) and Pearson's r (B) coefficients of correlation between knee flexion angle and body mass or lower limb length at gait events, with p-values indicated. Correlations significant at p < 0.05 are bolded. Note that the coefficient in the sample of Gruss (2007) at event V3 has a negative sign, in contrast to the published value. This is due to the reversed definition of knee flexion angle. (For definition of gait event abbreviations, see Fig. 2 caption.)*

		Gait events						
		IC	V1	V2	V3	TOFF	K1	K2
A.	Our sample (n = 26)[1]							
	Body mass	−0.01	−0.02	−0.04	−0.25	0.03	0.09	−0.10
		p = 0.930	p = 0.888	p = 0.800	p = 0.108	p = 0.825	p = 0.536	p = 0.453
	Lower limb length	0.01	0.05	−0.03	−0.19	0.06	0.16	−0.06
		p = 0.965	p = 0.735	p = 0.865	p = 0.214	p = 0.691	p = 0.261	p = 0.643
	Gruss (2007) (n = 27)							
	Lower limb length	-	x	x	**−0.29**	-	-	-
		-	ns	ns	**p < 0.05**	-	-	-
B.	Our sample (n = 26)[1]							
	Body mass	−0.04		0.02	**−0.44**	0.14	0.11	−0.17
		p = 0.851	non-normal	p = 0.945	**p = 0.038**	p = 0.542	p = 0.597	p = 0.402
	Lower limb length	0.00	distribution	−0.01	−0.34	0.14	0.16	−0.14
		p = 0.990		p = 0.975	p = 0.117	p = 0.526	p = 0.441	p = 0.493

[1] At events V1-3 the sample size is reduced to n = 22.
ns, non-significant correlation; -, data not available; x, specific value not provided.

length, the strongest correlation coefficients being −0.30 at 71 % of stance for body mass and −0.27 at 66 % of stance for lower limb length. The strongest negative correlations in late stance are close to the usual timing of event V3, and especially for correlation between knee flexion angle and body mass.

Linear regressions of knee flexion angle on body mass and lower limb length at event V3 are presented in Fig. 5. Event V3 was chosen for further examination because we detected significant correlation between knee flexion angle and body mass at this event and because in the previous study by Gruss (2007) significant correlation was detected at this event also between knee flexion angle and lower limb length. The slope of the linear regression of knee flexion angle on body mass is −0.092. According to the equation for the linear regression of knee flexion angle on body mass, an individual with 55 kg of body mass would be expected to have 6.0° knee flexion at V3, whereas an individual with body mass 95 kg would be expected to have 2.3° knee flexion at event V3. The slope of the linear regression of knee flexion angle on lower limb length is not significantly different from zero.

Discussion

Our results support the hypothesis that body size influences limb posture on the intraspecific level of variation. We found in humans significant negative correlation between body mass and knee flexion angle at the second peak of vertical GRF, which suggests that individuals with greater body mass keep their knees more extended at this gait event. This finding was strengthened by detection of strengthened although statistically insignificant negative correlation between knee flexion angle and body mass between 50 % and 90 % of normalized stance phase with the strongest correlation at 71 % of stance, which is close to the usual occurrence of the second peak of vertical GRF. We found no significant correlation between knee flexion angle and lower limb length at any gait event or at any percentage point of normalized stance phase. Although not statistically significant, strengthened negative correlation between knee flexion angle and lower limb length was, however, detected in late stance, and this correlation coincides almost exactly with strengthened negative correlation between knee flexion angle and body mass.

The insignificance detected in our study for the relationship between lower limb length and knee flexion angle during walking (Table 3) is inconsistent with previous findings of Gruss (2007), who found in her human sample significant negative correlation between knee flexion angle and lower limb length at the second peak of vertical ground reaction force and at the peak of propulsive ground reaction force. We cannot exclude that differences between the results of the present study and those of the Gruss (2007) study could be caused by differences in sample characteristics, in particular by differences in body proportions. Our sample has significantly lower mean BMI than did the sample of Gruss (2007), mostly due to the high proportion of overweight individuals included in Gruss' sample (11 overweight and one obese individual vs. six overweight and no obese individuals in our sample). Furthermore, mean lower limb length was greater in our sample compared to Gruss' sample. Although, some of the difference in lower limb length is due to the insignificant difference in mean stature and a lesser part is due to a different definition of the distal end of the shank (we measured to the distal tip of the *malleolus*, whereas Gruss (2007) measured to the medially most-projecting point on the *malleolus*), we propose that individuals in our sample had relatively longer lower limbs and thus shorter trunks. Differences in the marker set used to track kinematics might contribute to differences in magnitudes of knee flexion angle (Perry & Burnfield 2010), and this might have influenced the significance of the relationship between knee flexion angle and lower limb length. Mean knee flexion angles at particular gait events were higher in Gruss' sample than in our sample (Table 2, Fig. 3). However, Gruss' values are considerably greater even when compared to values from other studies (Murray 1967, Kadaba et al. 1990, Benedetti et al. 1998, Nordin & Frankel 2001, Lay et al. 2006).

It has been shown that lower limb posture influences moments and thus also the forces acting on bones and muscles through changes in length of the moment arm of the GRF (Biewener 1983, 1989a, Polk 2002, 2004, Gruss 2007, Reilly et al. 2007). It is believed that changes in limb posture can act as a moderating mechanism for larger individuals to keep the external moments low. In the human knee, such mechanism would be expected especially in early stance, when the external moment flexes the knee. More extended knee posture would shorten the moment arm of the GRF and thus lower the flexing moment. This mechanism has already been reported in studies comparing kinematics of lean and obese humans (DeVita & Hortobágyi 2003, Gushue et al. 2005). Obese individuals with greater body mass used less flexed knee postures in early stance of walking than did lean individuals and had knee flexing moments comparable to those of lean individuals despite their having twice the body mass. In the present study, however, we found no relationship between knee flexion angle and body size in early stance.

The absence of a relationship between knee flexion angle and body size in early stance is somewhat surprising, and we suggest two possible explanations. One possible explanation for this paradox could be found in the hypothesis of Browning & Kram (2007), that there might be some critical level of BMI or body mass above which individuals adjust their knee posture to reduce knee-joint loads. Individuals in our sample could have body mass and BMI under the hypothesized critical level. Second, knee flexion in early stance is also an important contributor to shock absorption of loading (Richards 2008, Perry & Burnfield 2010) and helps to moderate the peaks of the vertical GRF (Mochon & McMahon 1980). The need to control the peaks of the vertical GRF in early stance by keeping the knee flexed might offset the benefits of shortening the moment arm of the GRF by using more extended knee postures.

Interpretation of the relationship between knee flexion angle and body mass in late stance through knee moments is more complicated due to the great variability of reported moments between measuring protocols and even among individuals measured by the same technique. Although most studies have reported that the external moment in late stance, and particularly at the phase where the V3 event most likely occurs, is (except for the end of the stance) usually extending the knee (Kirtley 2006, Lay et al. 2006, Richards 2008, Perry & Burnfield 2010), others have reported that it flexes the knee (Kadaba et al. 1989).

Gruss (2005, 2007) suggested that more extended knee postures in late stance shorten moment arms of the GRF and thus reduce the knee flexing moment and the bending moments of the femur and tibia. Her arguments are weakened, however, by the facts that most studies have reported there to be a knee extending moment in late stance and that such extending moment even is displayed in two individuals within Gruss' sample (Gruss 2007: Fig. 6, p. 110). If there is an extending moment present in late stance, then further knee extension in this period of the gait cycle should lead to an increased knee extending moment through elongating of the moment arm of the GRF. However, Gruss (2005, 2007) did not detect any relationship between lower limb length and knee moment arm of the GRF or knee moment in late stance, and this could be caused by conversion of these variables to absolute values prior to correlation analysis. Even if there would be a relationship between knee moment or moment arm and lower limb length (the shorter-limbed individuals having a flexing moment and longer-limbed individuals having an extending moment), it would not be possible to detect this using

Gruss' approach. At late stance, a knee extending moment may be favourable to counterbalance the activity of the *gastrocnemius*, which plantar flexes the ankle with the aim to prepare the foot for toe off. However, we consider the benefit of increased knee extension in the late stance to be rather modest, if it exists at all. From another point of view, increased knee extension in late stance might actually be a consequence of the increased knee extending moment caused by greater body mass.

Conclusion

In summary, body mass is negatively correlated with knee flexion angle at the second peak of the vertical GRF in humans ($r = 0.444$, $p = 0.038$). No significant correlation was identified between lower limb length and knee flexion angle at any gait event. Although not significant, strengthened negative correlation between knee flexion angle and both body mass and lower limb length was detected between 50 % and 90 % of normalized stance phase with the maxima at 71 % of stance phase for body mass and 66 % for lower limb length. Thus, our results support the view that body size influences limb posture during locomotion even on the level of intraspecific variability. Among humans in particular, individuals with greater body mass tend to use more extended knee postures in the late stance phase of walking.

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