How does normal variability in trunk flexion affect lower limb muscle activity during walking?

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Abstract:

A large proportion of the mass of the body is contained within the trunk segment. Therefore, small changes in the inclination of this segment have the potential to influence the direction of the ground reaction force and alter lower limb joint moments and muscle activation patterns during walking. The aim of this study was to investigate if variability in sagittal trunk inclination in healthy participants is associated with differences in lower limb biomechanics. Gait analysis data was collected on 41 healthy participants during walking. Two groups were defined based on habitual trunk flexion angle during normal walking, a forward lean group (n=18) and a backward lean group (n=17). Lower limb moments, muscle activation patterns and co-contraction levels were compared between the two groups using independent t-tests. The forward lean group walked with 5° more trunk flexion than the backward lean group. This difference was associated with a larger peak hip moment (effect size=0.7) and higher activation of the lateral gastrocnemius (effect size =0.6) and the biceps femoris (effect size =0.7) muscles. The forward lean group also exhibited greater co-contraction in late stance (effect size =0.7). This is the first study to demonstrate that small differences in trunk flexion are associated with pronounced alterations in the activation of the lateral knee flexor muscles. This is important because people with knee osteoarthritis

have been observed to walk with increased trunk flexion. It is possible that increased sagittal trunk inclination may be associated with elevated joint loads in people with knee osteoarthritis.

Keywords: EMG, lower limb, trunk flexion, forward lean, co-contraction, knee osteoarthritis

1. Introduction

Cadaver studies have demonstrated that approximately 65% of the total mass of the body is contained within the head, arms and trunk (HAT) segment (Dempster, 1955). Therefore, small changes in the sagittal plane inclination (flexion) of the trunk have the potential to alter the position of the centre of mass relative to lower limb joint centres (Nordin & Frankel, 2001). Such changes will affect sagittal plane joint moments (Kluger, Major, Fatone, & Gard, 2014) and impact on muscle activation patterns (Grasso, Zago, & Lacquaniti, 2000). For example, an increase in trunk flexion will create an anterior shift of the centre of mass relative to the hip joint (Saha, Gard, & Fatone, 2008). This will lead to an increase in the internal hip extensor moment (Leteneur, Gillet, Sadeghi, Allard, & Barbier, 2009) which may lead to a corresponding increase in hip extensor muscle activity (Grasso et al., 2000). Given the relatively large mass of the HAT segment, small increases in trunk flexion have the potential to increase hamstring activity, through a change in hip moments, and gastrocnemius activity, through a change in ankle moments. As both these muscles also cross the knee joint, flexor-extensor co-contraction patterns at the knee may also be affected.

There is a substantial body of research which has demonstrated that people with knee osteoarthritis (OA) walk with altered muscle activation. These differences are typically characterised by prolonged and increased activity of the hamstrings (Hortobagyi et al., 2005; Rutherford, Hubley-Kozey, & Stanish, 2013) quadriceps (Hubley-Kozey, Deluzio, Landry, McNutt, & Stanish, 2006; Rutherford, Hubley-Kozey, Stanish, & Dunbar, 2011) and gastrocnemius (Astephen, Deluzio, Caldwell, Dunbar, & Hubley-Kozey, 2008; Childs, Sparto, Fitzgerald, Bizzini, & Irrgang, 2004) muscles. Importantly, other research has shown that these altered muscle patterns will increase the rate of cartilage loss (Hodges et al., 2016) and also increase the likelihood of having needed a total knee replacement at five-year follow up (Hatfield, Costello, Astephen Wilson, Stanish, & Hubley-Kozey, 2020). It has been suggested that disease-related mechanisms, such as increased active stiffness to counteract joint instability (Schipplein & Andriacchi, 1991), may underlie increased muscle activity in people with knee OA. However, it is also possible that increased trunk flexion may be the driver for increased muscle activation in people who have this disease. In a recent study, Preece et al. (2019) showed that people with knee osteoarthritis walk with a larger trunk flexion angle in comparison to healthy control participants. This finding motivates the need for research to explore whether small changes in trunk flexion could be associated with clinically important changes in knee muscle activation during walking.

To understand the effect of trunk flexion, it is important to study healthy people who are unaffected by diseases, such as knee OA. In line with this idea, previous investigators have used repeated measures approaches to understand the effect of increasing trunk flexion in healthy people. For example Saha *et al.* (2008), showed both knee flexion and ankle dorsiflexion to increase when subjects were instructed to walk with a 25° and a 50° increase in trunk flexion. Using a similar approach, Kluger *et al.* (2014) observed a decrease in the ankle plantarflexor moment and an increase in the hip extensor moment as trunk flexion was increased. In another study, Grasso *et al.* (2000) published preliminary data on five participants suggesting that large changes in trunk flexion (approximately 50°) would substantially increase lower limb muscle activity during walking. Taken together these studies show that relatively large changes in trunk flexion can bring about changes in lower limb kinematics, moments and muscle activation patterns. However, each of these previous studies investigated the effect of large changes in trunk flexion. Therefore, they do not provide insight into the potential influence of small differences in trunk flexion, more typical of the differences between people with knee OA and healthy control participants.

Another approach which can be used to understand the effect of trunk flexion is to compare biomechanical characteristics between healthy subjects, grouped according to their habitual trunk flexion angle in walking. In line with this idea, Leardini *et al.* (2013) compared motions of the thorax and pelvis between two groups who differed by a mean of 7.5° in trunk flexion. This study showed that greater trunk flexion was associated with more thorax-to-pelvis motion, however they did not report on lower limb moments or muscle activation patterns. In another study, Leteneur *et al.* (2009) compared lower limb moments between two groups who differed by only 4.6° in trunk flexion. They observed that greater trunk flexion was associated with a prolonged hip extensor moment and a lower peak in the hip flexor moment. However, they did not measure EMG activity. Therefore, further research is required to understand whether small differences in trunk flexion are associated with differences in muscle activation during walking. The aim of this study was to compare lower limb moments and muscle activation patterns between healthy participants, grouped according to their habitual trunk flexion angle in walking. We hypothesised that a group with greater trunk flexion would exhibit larger hip extensor and ankle plantarflexor moments along with greater activity of the hamstrings and gastrocnemius muscles.

2. Methods

A total of 41 adult healthy participants (23 male) were recruited for this study. Subjects were excluded if they had suffered with lower limb pain or back pain within the last 6 months or if they had been diagnosed with any neurological disease. The mean (SD) age of the participants was 28 (12) years old, mass 70.7 (12) kg, height 1.71 (0.07) m and BMI 24.4 (4.6) kgm^{-2}. All subjects provided written informed consent to participate in the study and ethical approval was obtained from the ethics committee at the University of Salford.

A biomechanical walking assessment was carried out for each participant. For this assessment, kinematic and kinetic data were collected using an Oqus camera system (Qualisys, Sweden) (100 Hz) with two AMTI force plate (1500Hz) embedded in the walkway. Skin-mounted reflective markers were used to track motions of the pelvis, trunk and also the thigh, shank and foot of one lower limb which was selected at random. Following the protocol suggested by Armand *et al.* (2014) for tracking the thorax, a trunk segment was defined using markers placed on the greater trochanters and acromions. This segment was tracked using markers placed on the jugular notch and the second and eighth thoracic vertebrae. The pelvic segment was defined using markers placed over the right and left anterior superior iliac spines and the right and left posterior superior iliac spines and tracked with a rigid cluster of 3 markers positioned over the sacrum. Two rigid clusters of 4 markers were also used to track the motions of the thigh and shank and a system of 4 markers, placed over anatomical landmarks, used to track motion of the foot (Jones, Chapman, Parkes, Forsythe, & Felson, 2015). Ankle and knee joint centres were calculated as midpoints between the malleoli and femoral epicondyles respectively and hip joint centres obtained using the regression model of Bell *et al.* (1989).

Surface electromyography (EMG) data were collected from the hamstrings, quadriceps and gastrocnemius muscles using a Noraxon DTS system (1500 Hz). These data were obtained from the same limb selected for the kinematic/kinetic data. Following skin preparation with an abrasive gel and alcohol wipe, electrodes were placed over the vastus lateralis, vastus medialis, biceps femoris, semitendinosus, medial gastrocnemius and lateral gastrocnemius muscles according to SENIAM guidelines (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). Once electrodes were positioned, subjects were instructed to walk barefoot at a self-selected speed along a 6 metre walkway. Walking speed was measured using optical timing gates and only successful trials, when the foot was within the boundary of the force platform, accepted. A minimum of seven successful walking trials were recorded for each participant.

Following the walking trials, reference data from a maximum voluntary isometric contraction (MVIC) were collected for each muscle following a protocol suggested by Rutherford *et al.* (2011). With this protocol, the hamstring muscles were contracted with participants lying prone with their knee in 55° of flexion and the quadriceps muscles contracted with participants seated with the knee in 15° of flexion. To test the gastrocnemius muscles, participants stood on tip toes of the leg being tested. For the hamstrings and quadriceps tests, participants were required to maximally contract their muscle against fixed resistance, whereas for the gastrocnemius muscles participants were instructed to push upwards as hard as possible. For each test, subjects were given verbal encouragement and data collected for 5 seconds with a 60 second rest between contractions. Three separate tests were recorded for each of the three muscle groups.

Following data collection, marker data and force data were low pass filtered at 12Hz and 25Hz respectively. Lower limb joint kinematics and moments were then derived using a six degree of freedom model implemented using the Visual 3D software (C-Motion, Rockville, Maryland). The angle of the trunk segment was calculated relative to the laboratory system, such that trunk flexion angle represented the angle between the laboratory vertical and the trunk vertical axis (see segment definition above). Kinematic trajectories of the hip, knee and ankle were calculated as relative angles between adjacent segments using an X-Y-Z Euler rotation sequence. For the lower limb joint moment calculations, inertial properties of each segment were defined using the data of Dempster (1955) and joint moments calculated using the Visual 3D inverse dynamic algorithm. Hip, knee and ankle moments were then normalised by participant's body mass. Gait events were calculated by applying a

20N threshold to the vertical ground reaction force data and subsequently used to time normalise kinematic data to a full gait cycle and time normalise the kinetic data to stance phase. Ensemble average data for each kinematic/kinetic trajectory was then calculated for each separate participant across all walking trials. This ensemble averaging was repeated for each kinematic/kinetic trajectory to create the final dataset.

The dynamic EMG was high pass filtered (20 Hz) to remove movement artefact and then rectified. A linear envelope was then created using a 6 Hz low pass Butterworth filter. Similar to the joint moment data, gait events were used to time normalise the EMG data to stance phase and ensemble average curves created for each muscle. The MVIC data was then high pass filtered (20 Hz), rectified and a linear envelop created. Following the approach proposed by Hubley-Kozey *et al.* (2006), a 0.1 s moving window algorithm was used to determine an MVIC for each muscle from each trial. The maximal value, across the three MVIC trials, was then used to normalise the dynamic EMG data. In addition to the normalised muscle activation profiles for the six individual muscles, four co-contraction activation profiles were derived. The first two profiles were obtained by summing medial/lateral hamstring and quadriceps activity and the second two profiles obtained by summing medial/lateral gastrocnemius and quadriceps activity. Previous authors have advocated using a specific co-contraction ratio (Heiden, Lloyd, & Ackland, 2009). However, we chose to sum medial knee flexor and knee extensor activity as modelling studies have shown that this approach is more closely related to joint contact forces than the co-contraction ratio (Winby, Gerus, Kirk, & Lloyd, 2013). It was felt that this approach would provide insight into the potential influence of trunk flexion on joint loading.

Subjects were divided into two groups based on their habitual trunk flexion angle in walking. Specifically, trunk flexion was averaged across the gait cycle for each participant and the median trunk flexion angle, across all participants, identified. We used a modified median split approach in which we excluded participants who exhibited a trunk angle which was within a region of uncertainty of the median. The region of uncertainty was identified from test-retest data from five participant (tested on two occasions separated by one week) which showed a standard error of measurement of 0.9^{o}. Six participants were found to lie within 0.9^{o} of the median trunk angle and were therefore not allocated to either of the two groups. The remaining 35 participants were defined as either forward leaners (FW), n=18, or backward leaners (BW), n=17, depending on whether their habitual trunk flexion was above or below the median trunk angle.

Mean trunk flexion during walking was 4.8^{o} in the FW group (range 3.3° to 7.1°) and -0.2^{o} in the BW group (range 1.5^{o} to -3.7^{o}). Additional demographic and spatiotemporal parameters are show in Table 1. Independent t-tests, or Mann-Whitney tests for non-parametric data, were performed on these data and confirmed that there were no differences between the two groups apart from trunk flexion. To investigate differences in hip and ankle joint moments between the groups, the following outcome variables were derived: peak hip extensor moment and peak ankle plantarflexor moment. These peaks were chosen to facilitate comparison with previous work investigating differences in trunk flexion (Leteneur et al., 2009). To investigate differences in hamstring and gastrocnemius muscle activation between the two groups, the mean timing (across all 35 participants) of peak muscle activity was first identified for each of the four separate muscles. This timing of peak activity was used to centre a window, width 20% of stance phase, for each muscle. Normalised EMG data was then averaged across the corresponding window to derive a measure of muscle activation. To investigate differences in co-contraction we defined windows which were centred on the timing of peak joint contact loads (Brandon, Miller, Thelen, & Deluzio, 2014). In their modelling study, Brandon *et al*. (2014) reported peaks in the axial knee contact force at 13% and 45% of stance phase which equates to 20% and 75% of stance phase. We therefore selected windows across 10-30% stance phase for hamstring-quadriceps co-contraction and 65-85% of stance phase for gastrocnemius-quadriceps co-contraction.

TABLE 1 HERE

Moments, muscle activation and co-contraction were compared between the FW and BW groups using independent t-tests as all data were normally distributed. To control for type I error, we used a Bonferroni-Holm correction to adjust the critical α of 0.05. This correction was applied separately to the two joint moment outcomes, the four individual muscle activation outcomes and the four co-contraction outcomes. We also used Cohen's d, effect size (ES), to quantify the magnitude of the difference between the two groups.

3. Results

There was a pronounced difference in mean trunk inclination between the two groups, however, the temporal profile for each group was very similar (Figure 1a). The mean difference of 5^{o} in trunk flexion between the groups was associated with a 28% larger hip extensor moment in the FW group (Table 2). Although there were subtle differences in the profile of the ankle and knee moments (Figure 1c & 1d), there were no significant differences in ankle moments (Table 2). We did not perform statistical analysis on the knee moment as our specific hypotheses were related to the hip and ankle moments.

TABLE 2 & FIGURE 1 HERE

Pronounced differences in lateral muscle activity were observed between the FW and BW groups (Figure 2). The FW group exhibited 33% more biceps femoris activity and 45% more lateral gastrocnemius activity across the periods of stance phase corresponding to peak activity (Table 3). Interestingly, although the FW group showed higher lateral hamstring activity for the whole of early stance, there were minimal differences in the temporal profile of the medial hamstring (Figure 2). Compared to the BW group, the FW group appeared to show higher activation of medial gastrocnemius in late stance, however, this difference did not reach statistical significance across the period corresponding to peak activity (Table 3). We did not perform statistical analysis on knee muscle activations as our specific hypotheses were related to hamstring and gastrocnemius activity.

TABLE 3 & FIGURE 2 HERE

Co-contraction of the lateral knee flexor-extensor muscles appeared to be larger in the FW group compared to the BW group (Figure 3). This difference reached statistical significance, over the period corresponding to peak loading, for lateral gastrocnemius-quadriceps co-contraction (effect size = 0.72, Table 4). However, this difference was less pronounced for medial gastrocnemius-quadriceps co-contraction and did not reach statistical significance (Table 4).

FIGURE 3 HERE

4. Discussion

The aim of this study was to investigate differences in sagittal hip and ankle moments, muscle activation and co-contraction patterns between groups of healthy subjects who habitually walk with different trunk flexion angles. In line with our hypotheses, the data indicate that walking with increased trunk flexion is associated with a greater hip extensor moment, higher activity of the lateral knee flexor muscles and with greater gastrocnemius-quadriceps co-contraction in late stance. However, we did not observe a difference in ankle plantarflexor moment between the groups. Nevertheless, the results support the idea that relatively small differences in sagittal plane inclination of the trunk during walking appear to be associated with pronounced differences in muscle activation.

Our data on joint moments are generally consistent with the observations of Leteneur *et al.* (2009) who divided subjects into groups of FW and BW which differed by a 4.6°. Although they did not observe a difference in peak moments, they did observe a more prolonged hip extensor moment in the FW group, an observation which appears consistent with our data (Figure 1b). Similar to our findings, Leteneur *et al.* (2009), did not observe a difference in ankle moments, however they did not report on EMG activity. Our observation of higher hamstring activity is consistent with the observation of an increased and/or prolonged hip extensor pattern. However, it is unclear why this was observed in the lateral but not the medial hamstring (Figure 2). Interestingly, our data, and data of Leteneur *et al.* (2009), do not support an association between increased trunk flexion and an increased ankle plantarflexor moment. However, we did observe a higher level of activation of the lateral gastrocnemius. This inconsistency could be the result of increased ankle dorsiflexor muscle activity, e.g. tibialis anterior, which would lead to increased co-contraction at the ankle but no net increase in joint moment. Clearly, additional EMG measurements would be required to investigate this idea further.

We observed a range of trunk flexion angles from -4° to 7° across our cohort of 35 healthy participants. This range appears consistent with previous reports of trunk flexion in healthy people (Turcot et al., 2013) and with the data reported by Leteneur *et al.* (2009). However, it is difficult to make precise comparisons because of the different anatomical coordinate systems which have been used by previous authors. Nevertheless, it would appear that a range of approximately 10° is typical for healthy people and this is likely to go some way towards explaining the variability in hip extensor (Winter, 1984) and hamstring patterns (Winter, 1991) observed in healthy people. For example, previous research has observed that increased hip extensor moments are associated with older age (DeVita & Hortobagyi, 2000). It is possible that this observation is the result of differences in trunk flexion across different age groups.

People with knee OA have been observed to walk with increased and prolonged activation of the hamstrings (Hortobagyi et al., 2005; Rutherford et al., 2013), prolonged activation of gastrocnemius muscles (Childs et al., 2004) and increases in muscle co-contraction (Heiden et al., 2009; Hortobagyi et al., 2005; Preece, Jones, Brown, Cacciatore, & Jones, 2016). Such alterations in muscle activation have been shown to increase compressive force at the knee joint (Brandon et al., 2014) and to increase the rate of cartilage loss (Hodges et al., 2016). It has been suggested that this increased muscle activation may act to stabilise the knee joint (Rutherford et al., 2013). However, our data suggests a link between trunk flexion and both increased lateral hamstring activation and increased gastrocnemius-quadriceps co-contraction in people who do not have pain. Although our data provide no insight into cause or consequence of this association, it is possible that previously observed alterations in muscle activity, known to increase joint loading (Brandon et al., 2014), may be related to increased trunk flexion during walking.

In a recent study, Preece *et al.* (2019) observed that people with knee OA walk with 2.6° more trunk flexion than healthy control participants. Although this is less than the 5° difference between the FW and BW groups reported in this current study, it does support the possibility that increased trunk flexion may explain some of the alterations in muscle activity in people with knee OA. For example, Hubley-Kozey *et al.* (2009) and Rutherford *et al.* (2013) observed a profile of lateral hamstring activation, characterised by increased activation at initial contact and during the first half of stance. This pattern is very similar to the pattern observed in the FW group in this current study (Figure 2). Similarly, Childs *et al.* (2004) observed an earlier onset of medial gastrocnemius in people with knee OA, a pattern which appears qualitatively similar to the pattern exhibited by the FW group in this current study (Figure 2a). Given these similarities, further research is required to quantify the degree to which observed differences in muscle activation, between people with knee OA and controls, may be explained by variation in trunk flexion angle.

There are number of limitations to the present study which should be highlighted. Firstly, normalisation of EMG signals by MVIC can be problematic, especially if a comparison is being performed on participants who find it difficult to fully activate their muscles. However, there were no demographic differences between the FW and BW groups in this study. Therefore, it is unlikely that there could have been differences in the MIVC data between the two groups sufficient to distort group comparison. Another difficulty relates to quantification of the trunk which is a multi-articulate structure, and which therefore must be segmented into pseudo-rigid segments for kinematic analysis. Although our modelling approach, in which we defined a single rigid trunk segment, does not provide insight into intervertebral motions, it can be used to provide an overall measure of trunk inclination. Furthermore, we used a protocol which has been shown to be optimal for quantifying trunk motion (Armand et al., 2014) and which we have successfully used before to demonstrate differences in trunk flexion between healthy people and people with knee OA (Preece et al., 2019).

This is the first study to investigate how small differences in sagittal plane inclination of the trunk could impact on lower limb muscle activation patterns. Our data show a greater activation of the lateral knee flexor muscles and higher gastrocnemius-quadriceps co-contraction in healthy people who walk with increased trunk flexion. These patterns of altered muscle activity have been observed in people with knee OA and are known to increase joint loading. It is therefore important to understand whether interventions, which can decrease trunk flexion in people with knee OA, could bring about reductions in knee flexor activity and corresponding reductions in joint loading.

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Tables:

Table 1: Demographic and spatiotemporal parameters for the backward lean (BW) group and the forward lean (FW) group.

	Backward lean (BW)	Forward lean (FW)
Number of subjects	17	18
Gender	10 Male 7 Female	12 Male 6 Female
Age (Years)	36 (11)	34 (13)
BMI (kgm^{-2})	25.3 (5.2)	23.6 (3.8)
Speed (ms^{-1})	1.27 (.14)	1.31 (.13)
Step length (m)	1.29 (0.08)	1.32 (.10)
Limb	10 R leg and 7 L leg	12 R leg and 6 L leg

Table 2: Mean (SD) trunk flexion and hip and ankle moments for the backward lean (BW) group and the forward lean (FW) group. The hip and ankle moment variables are presented in rank order according to p-value and significant differences (Bonferroni–Holm method used to adjust the critical alpha of 0.05) denotated by *. Non-parametric testing was used to investigate trunk angle and therefore t-test statistics have not been reported.

	BW Mean (SD)	FW Mean (SD)	P-value	Effect size	t(df)
Trunk angle (°)	-0.2 (1.5)	4.8 (1.2)	**<0.001***	**2.5**	**-**
Hip moment ($Nmkg^{-1}$)	.54 (.15)	.69 (.23)	**0.02496***	**0.78**	**2.349(33)**
Ankle moment ($Nmkg^{-1}$)	1.45 (.16)	1.47 (.13)	0.68	0.14	0.41(33)

Table 3: Mean (SD) muscle activity for the backward lean (BW) group and the forward lean (FW) group for each of the four knee flexor muscles. The variables are presented in rank order according to p-value and significant differences (Bonferroni–Holm method used to adjust the critical alpha of 0.05) denotated by *.

Muscle	Time window (% Stance)	BW Mean (SD) EMG activity (% MVIC)	FW Mean (SD) EMG activity (% MVIC)	P-value	Effect size	t(df)
Biceps Femoris	-20 – 0	12 (2)	16 (6)	**0.01***	**0.66**	**2.71(33)**
Lateral Gastrocnemius	55-75	31 (10)	45 (22)	**0.016***	**0.57**	**2.53(33)**
Semitendinosus	-20 – 0	15 (9)	18 (9)	0.42	0.19	0.80(33)
Medial Gastrocnemius	50-70	43 (10)	45 (18)	0.75	0.07	0.32(33)

Table 4: Mean (SD) co-contraction for the backward lean (BW) group and the forward lean (FW). The variables are presented in rank order according to p-value and significant differences (Bonferroni–Holm method used to adjust the critical alpha of 0.05) denotated by *.

Muscle pair	Time window (% Stance)	BW Mean (SD) EMG activity (% MVIC)	FW Mean (SD) EMG activity (% MVIC)	P-value	Effect size	t(df)
Lateral Gastrocnemius – Vastus Lateralis	65-85	24 (9)	41 (21)	**0.006**	**0.72**	**2.97(33)**
Medial Gastrocnemius - Vastus Medialis	65-85	27 (10)	35 (17)	0.08	0.43	1.78(33)
Biceps Femoris - Vastus Lateralis	10–30	22 (9)	28 (21)	0.30	0.26	1.06(33)
Semitendinosus - Vastus Medialis	10–30	21 (9)	23 (14)	0.61	0.12	0.51(33)

Figures

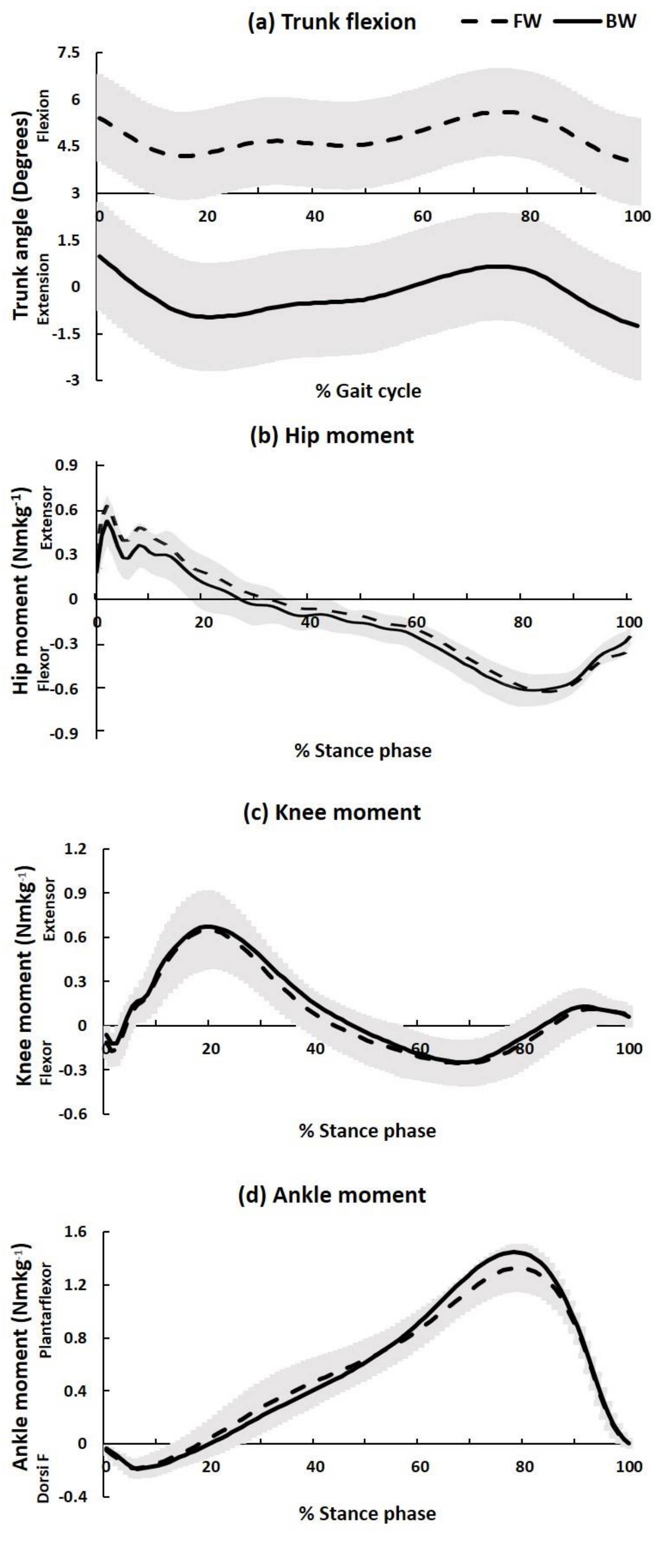


Figure 1: Ensemble average profiles for trunk flexion, hip moment, knee moment and ankle moment for the forward lean (FW) and the backward lean (BW) lean groups. The shaded area indicates the SD of the BW group.

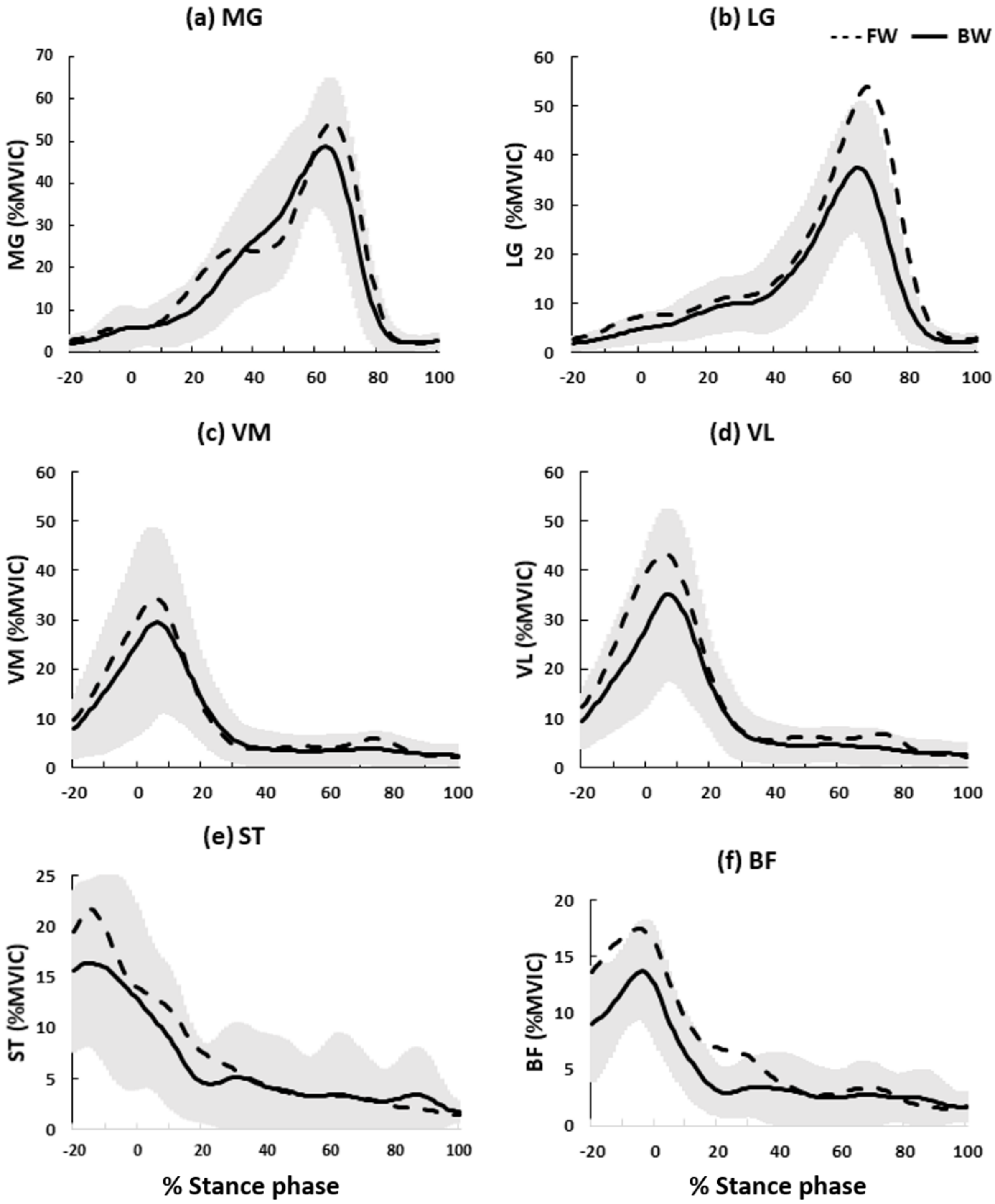
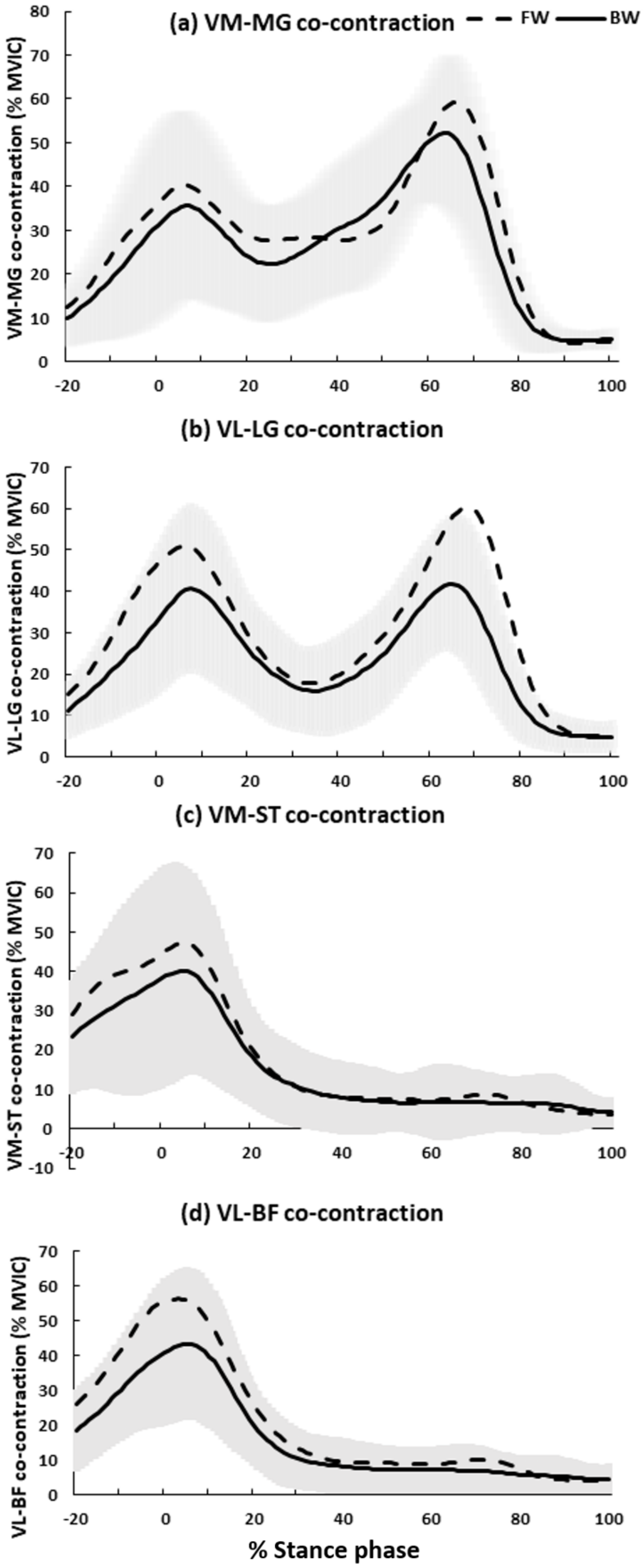


Figure 2: Ensemble average profiles for muscle activation for the forward lean (FW) and backward lean (BW) lean groups. Data is shown for medial/lateral gastrocnemius (MG, LG), vastus medialis/ lateralis (VM, VL), semitendinosus (SM) and biceps femoris (BF). The shaded area indicates the SD of the BW group.



501 Figure 3: Ensemble average profiles for co-contraction for the forward lean (FW) and backward lean (BW) lean groups. The shaded area indicates the SD of the BW group.