LOW IMPACT WEIGHT-BEARING EXERCISE IN AN UPRIGHT POSTURE ACHIEVES GREATER LUMBOPELVIC STABILITY THAN OVERGROUND WALKING

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ABSTRACT

The aim of this study was to determine the kinematic differences between movements on a new exercise device (EX) that promotes a stable trunk over a moving, unstable base of support, and overground walking (OW). Sixteen male participants performed EX and OW trials while their movements were tracked using a 3D motion capture system. Trunk and pelvis range of motion (ROM) were similar between EX and OW in the sagittal and frontal planes, and reduced for EX in the transverse plane. The pelvis was tilted anteriorly, on average, by about 16 degrees in EX compared to OW. Hip and knee ROM were reduced in EX compared to OW. The exercise device appears to promote similar or reduced lumbopelvic motion, compared to walking, which could contribute to more tonic activity of the local lumbopelvic musculature.

Keywords: kinematics, walking, lumbopelvic stability, exercise

INTRODUCTION

In vitro studies have shown the thoracolumbar and lumbar spine, devoid of any musculature, will experience structural failure under compressive loadings as small as 20 and 90 N in magnitude, respectively (Crisco et al 1992). Considering spinal loadings experienced *in vivo* can range from 6 kN during selected everyday tasks (McGill & Norman 1986) to in excess of 36 kN during competitive powerlifting (Cholewicki et al 1991) the human vertebral column is intrinsically incapable of meeting the physiological demands placed upon it without additional stabilisation at a segmental level (Panjabi et al 1989).

The role of the lumbopelvic trunk musculature in providing the required supplementary stability at a segmental level is well documented (Bergmark 1989; Panjabi 1992; Cholewicki & McGill 1996; Vera-Garcia et al 2007). In particular, due to their anatomical positioning, morphology and function, the deeper fibres of the lumbar multifidus (LM) and the transversus abdominis (TrA) are considered crucial for local stability of the lumbar spine (Hodges & Richardson 1996; Hodges 1999; Kim et al 2007).

A growing body of evidence links structural and functional changes of local stabilising trunk muscles with low back pain (LBP) (Hides et al 1994; Hides et al 1996; Hodges & Richardson 1996; Danneels et al 2000; Oddsson & De Luca 2003; Hides et al 2008; Hides et al 2008; MacDonald et al 2009; Teyhen et al 2009; Wallwork et al 2009). In people with LBP, muscle fibre atrophy and fatty infiltrations of the LM have been observed (Kader et al 2000), as well as a dysfunction of the anticipatory activity of the LM and TrA (Hodges & Richardson 1998).

Corrective/restorative treatment strategies for such dysfunction of the local lumbopelvic musculature have included specific motor control exercises (Hides et al 2008), ‘core stability’ training, muscular strength and endurance training (Danneels et al 2001), aerobic exercise (Frost et al 1995) and the use of an unstable base of support (BOS) (Marshall & Murphy 2006), often in a tailored combination (Demoulin et al 2010). The majority of these approaches tend to show only modest effectiveness (Keller et al 2007; van Middelkoop et al 2010), possibly due to a lack of carry-over to functional day-to-day activities (Richardson & Hides 2004; Hodges & Cholewicki 2007).

Recently a new method promoting activation of LM and TrA has been proposed as an alternative to the current approaches for addressing local lumbopelvic muscle dysfunction (Debuse et al 2013). The users of the exercise device move their feet in a quasi-elliptical path in anti-phase against virtually no external resistance. The absence of external resistance creates the need for much greater motor control of the legs and pelvis, to control leg movement, whilst maintaining an upright trunk posture, than in conventional exercise devices. The exercise device was found to recruit LM and TrA to a greater extent than a range of control activities, including standing on the ground or on an unstable base of support and voluntary muscle contractions. The authors postulated that the method promotes a relatively stable lumbopelvic area during a functional lower limb movement and results in an automatic recruitment/activity of TrA and LM (Debuse et al 2013). Richardson and

Jull in their seminal paper of 1995 proposed that local muscles work tonically, as opposed to global muscles which tend to work phasically. This is widely accepted by other authors working in this field (for example Sahrmann 2002; Hides 2004; Hides et al 2004; Hodges & Cholewicki 2007). Debuse et al (2013) imply that tonic muscle activity is likely to be responsible for the stable lumbopelvic region when using the exercise device. However, no information was provided on the lumbopelvic and lower limb kinematics of the user while exercising to identify how the exercise device promoted lumbopelvic stability and, thus, tonic muscle activity.

The aim of the current study was to compare lower limb, pelvic and trunk kinematics during the use of a newly developed exercise device (EX) and overground walking (OW), with a particular focus on the level of lumbopelvic stability in both activities.

METHOD

Participants

Sixteen healthy adult male volunteers (mean ± SD age: 26.5 ± 3.38 years, body mass: 82.158 ± 7.21 kg, height: 1.78 ± 0.05 m, and body mass index: 25.89 ± 2.16 $kg \cdot m^{-2}$) with no recent history of LBP, gait impairments, or other conditions affecting their ability to walk or exercise, agreed to participate in this study. Participants gave their fully informed written consent to take part. The study had received ethical approval from the Institutional Review Board prior to data collection.

Three-dimensional Motion Capture

Three-dimensional trajectories of 39 retro-reflective markers (Ø=14mm) were captured at a sampling frequency of 200 Hz using a 12 camera near-infrared motion capture facility (MX T20, Vicon Motion Systems, Oxford, UK). Markers were placed in accordance with a standard full-body model (Plug-in-Gait, Vicon Motion Systems, Oxford, UK), which consists of a 15 segment rigid-linked model of the head, thorax, pelvis, and bilateral upper arms, forearms, hands, thighs, lower legs and feet. Only the segmental orientations of the thorax, pelvis, thighs and lower legs were subsequently used for analysis.

The motion capture system was calibrated before all testing sessions using a standard dynamic protocol, with a 5 marker calibration wand (Vicon Motion Systems, Oxford, UK). System calibration was accepted when the image error of all 12 cameras was less than 0.2 mm.

Body mass, height and anthropometric measurements, including leg length (anterior superior iliac spine to medial malleolus), ankle widths and knee widths, necessary for the correct operation of the model used were taken in triplicate and the mean value used thereafter.

Experimental protocol

Participants completed an overground walking (OW) condition and a condition using the exercise device (EX – Figure 1) in a counterbalanced random order within a single session. In the OW condition participants were asked to walk along a level 7.5 m walkway, instrumented with embedded force plates (OR6-7, AMTI, Watertown, Massachusetts, USA), at a self-selected comfortable speed. Starting positions were adjusted individually to ensure that ‘clean’ foot contacts with the force plates could be achieved without direct targeting by the participant. A minimum of 10 trials were completed, before six trials - without evidence of targeting - were selected for subsequent analysis.

In the EX condition participants were given an initial five minute period to familiarise themselves with the exercise device. Following this, 30 seconds of trajectory data were captured during exercise in standing. Subsequently, six cycles were chosen at random for analysis. All participants were given standardised instructions on the correct use of the device emphasising the need for a ‘slow controlled movement’ whilst maintaining ‘an upright posture’ during each cycle.

Data processing and reduction

Marker trajectories collected during OW and EX trials were reconstructed and processed within Vicon Nexus (1.7, Vicon Motion Systems, Oxford, UK). Lost or obscured trajectory segments were interpolated using a quintic-spline function for gaps less than or equal to 10 frames (0.05 s) or a pattern fill function for gaps less than 10 frames, which utilises the trajectory of a marker with a similar predicted displacement trajectory. Marker trajectories were then low pass filtered at 5 Hz using a fourth-order zero lag Butterworth filter (Saunders et al 2005).

Key “gait cycle” phases (stance and swing) were demarcated for both the OW and EX conditions using discrete gait cycle events. Heel strikes and toe offs during OW were detected using the vertical component of the ground reaction force obtained from the force plates embedded flush with the walkway surface at the centre of the calibrated capture volume. When using the new exercise device, the feet remain in contact with the foot plates at all times during both stance and swing phase. Therefore, data collected during EX were divided into a stance and swing phase based on the trajectory of a marker placed on the front corner of the foot plate: stance was defined as the most anterior to the most posterior foot plate position, and swing was from the most posterior to most anterior foot plate position.

Three-dimensional angular displacements for the trunk (thorax with respect to [wrt] pelvis), pelvis (wrt the room, rather than a relative position between body segments), hip (pelvis wrt thigh) and knee (thigh wrt lower leg) were time normalised to cycle duration in 2% increments (51 data points from 0-100%) for the right sided cycles of both OW and EX conditions. Angular range of motion (ROM) was calculated as the maximum minus the minimum joint angle achieved within one cycle. This was done for each of the six trials and averaged within each participant, and then between all participants in both conditions. The mean angular position of each segment or joint was determined as the average of each angle throughout the gait cycle for OW and EX. The difference in mean angular positions, or offset, between OW and EX was calculated. Data for each variable were checked for normality of distribution using Q-Q and box plots. For variables that were normally distributed, paired samples t-tests were used to compare ROM and mean angular position between conditions with significance set at $p < 0.05$. For variables that were not normally distributed, Wilcoxon signed rank tests were instead used. Confidence intervals (95%) were also calculated for each pairwise comparison. All statistical analyses were performed using SPSS (version 19).

RESULTS

Spatiotemporal characteristics

All spatiotemporal data were normally distributed. Statistically significant differences were observed in all six spatiotemporal parameters (Table 1). The EX condition was characterised by reduction in cadence (t=21.220, df=15, p<0.001), stride length (t=14.041, df=15, p<0.001), stride duration (t=26.380, df=15, p<0.001), speed (t=20.506, df=15, p<0.001), and effective stance phase (t=15.354, df=15, p<0.001) compared to those observed during OW. Step width was significantly greater in the EX condition compared to OW (t=2.662, df=15, p<0.05).

Kinematics

All angular ROM data were normally distributed with the exception of the hip in the transverse plane. Angular ROM was found to be similar between EX and OW conditions for the trunk in the sagittal (t=1.622, df=15, p=0.126) and frontal (t=1.203, df=15, p=0.248) planes, and was similar for the pelvis in the sagittal (t=1.607, df=15, p=0.129) and frontal (t=0.213, df=15, p=0.834) planes. In the transverse plane, ROM was significantly reduced for the trunk (t=8.513, df=15, p<0.001) and the difference approached significance in the pelvis (t=1.854, df=15, p=0.083) between EX and OW (Table 2).

All mean angular position data were normally distributed with the exception of the pelvis and hip in the transverse plane. The pelvis was significantly tilted anteriorly for the EX condition compared to OW with an offset of 6.49° (t=4.697, df=15 p<0.001) (Table 3). Hip ROM was significantly reduced in the EX condition compared to OW in the sagittal (t=7.359, df=15, p<0.001), frontal (t=2.572, df=15, p=0.021) and transverse (Z=3.516, p<0.001) planes (Table 2). Knee ROM was also reduced in EX in the sagittal (t=8.463, df=15, p<0.001), frontal (t=7.041, df=15, p<0.001) and transverse (t=7.120, df=15, p<0.001) planes. The hip (t=13.297, df=15, p<0.001) and knee (t=19.878, df=15, p<0.001) were both more flexed throughout the gait cycle in the EX condition than in OW, with offsets of 22.31° and 24.11°, respectively, which were significant (Table 3). Despite the reduced ROM, peak knee and hip angles occurred at a similar point in the gait cycle for OW and EX (Figure 2).

DISCUSSION

The aim of this investigation was to compare the kinematics of lower limb and trunk motion during the use of a newly developed exercise device (EX), and overground walking (OW). The key findings of this study were that the lumbopelvic region was at least as stable whilst exercising on the new exercise device as overground walking. In the transverse plane, reduced ROM was observed during EX compared to OW. This stable lumbopelvic region was achieved over a dynamically moving base of support, where the ROM of the knees and hips was lower in EX than in OW. All spatiotemporal variables were significantly reduced in EX compared to OW, suggesting a slower, more controlled motion.

Trunk motion in the sagittal and frontal planes demonstrated similar ranges for both EX and OW. In the transverse plane, a reduced ROM was observed for EX suggesting increased lumbopelvic stability. Similar observations were made for the pelvis in terms of ROM, although in the transverse plane, a smaller reduction in range of motion was found for EX, with this reduction approaching statistical significance.

As a fundamental human activity, walking has previously been investigated as an intervention strategy in the treatment of LBP (Torstensen et al 1998; Joffe et al 2002; Taylor et al 2003; Mirovsky et al 2006). However, heterogeneity of study design and methodological quality have contributed to inconsistent findings (Hendrick et al 2010). Of these studies only Torstensen et al. (1998) and Taylor et al. (2003) used walking independently, while Joffe et al. (2002) and Mirovsky et al. (2006) combined walking with bodyweight support and traction, respectively. Notwithstanding the lack of evidence supporting walking as an effective intervention strategy for low back pain, the movement itself, involving control of trunk and pelvis motion during lower limb movements, is known to contribute to recruitment of the TrA and LM (Saunders et al 2004; Saunders et al 2005). Importantly, walking tends to be advocated by health care professionals in line with recommendations that ordinary physical activities should be continued as much as possible in order to aid recovery from LBP and prevent long-term disability (van Tulder et al 2000).

Similarities observed in both trunk and pelvic ROM between EX and OW in the sagittal and frontal planes suggest that the exercise device may be similar to walking, in terms of enabling tonic recruitment of the local lumbopelvic muscles such as TrA and LM. Previously Saunders et al. (2004; 2005) reported tonic TrA but phasic LM activity at walking speeds comparable to those reported here. However, no data were presented describing changes in activity amplitude, if any, within each gait cycle. The phasic activity of LM previously reported during walking (Saunders et al 2004) could be a factor leading to the questionable effectiveness of walking as a successful intervention for LBP (Hendrick et al 2010). The reduced transverse ROM, and thus the inherently more tonic muscle actions, in EX compared with OW seen in the current study could further indicate facilitation of greater tonic activity of the local lumbopelvic muscles (Richardson & Jull 1995) when using the new exercise device than in overground walking. If this reduced axial rotation results in more tonic recruitment of LM at a segmental level, then this could lead to the exercise device being a more successful intervention for LBP than walking. Current research within our group is exploring differences in lumbopelvic muscle recruitment between the exercise device and walking using ultrasound imaging and electromyography. Future studies in symptomatic populations are required to examine the clinical effectiveness of the exercise device.

No angular offsets were found between EX and OW for the trunk or pelvic position in all three planes, with the exception of a greater degree of anterior tilt of the pelvis in the EX condition. Influences of anterior pelvic tilt (O'Sullivan et al 2006) and accompanying lordotic spinal posture (Claus et al 2009), similar in magnitude to that observed within this investigation, have previously been shown to recruit both the superficial and deep fibres of the LM to approximately 30-40% of maximal voluntary isometric contraction capabilities, which is known to facilitate stabiliser muscle recruitment (McArdle et al 1991). Thus, this angular offset could be beneficial for the recruitment of the LM, provided care is taken to avoid over-recruitment of the superficial fibres of LM.

Hip and knee joints were more flexed throughout the gait cycle in EX than during OW. The increase in hip flexion was partly due to the angular definition being relative to a perpendicular axis of the pelvis. Therefore, the observed increase in anterior tilt creates a greater degree of flexion at the hip. The increased flexion of the knee throughout the gait cycle during EX are linked to the reduced stride length that was caused by the mechanical constraints of the device. By reducing stride length, the knee was unable to reach full extension during the stance phase of the gait cycle, as is normally seen during OW. What was apparent for knee and hip motion in the sagittal plane, was that the change in angle throughout the gait cycle showed a more sinusoidal pattern in EX compared to OW. This, more regular, movement pattern could contribute, to some extent, to more continuous/tonic muscle recruitment, a key training requirement of the local stabilising musculature (Richardson & Jull 1995).

There has been a drive, in recent years, for training interventions for the local muscles of the lumbopelvic region to be made more functional (Hodges 2011). A number of studies have brought into question the transferability of any training effects seen following less functional activities such as gym ball training where the base of support is simply unstable (Drake et al 2006). Debuse et al. (2013) demonstrated that the local lumbopelvic muscles were recruited to a greater extent with lower limb movement and an unstable base of support than with standing still on an unstable base of support (i.e. no voluntary lower limb movement). While overground walking involves lower limb movement, it does not usually involve an unstable base of support. During exercising on the new device, the requirement to control the descent of the "front" leg by gradually unloading the "back" leg within each gait cycle may result in greater recruitment of the local lumbopelvic muscles than overground walking. Our ultrasound imaging studies will examine this aspect in greater detail.

This study has a number of limitations. It examined relative motion between the pelvis and trunk. In order to gain a better understanding of how the exercise device influences the kinematics of the lumbopelvic region, a more detailed model of the thoracic and lumbar spine is needed. This would enable vertebral motion to be evaluated at a segmental level. Participants were asked to walk at their preferred walking speed. Due to the nature of the exercise device, movements were slower compared to walking. Saunders et al (2005) reported reduced axial rotation of the spine when walking slower. Thus, slow walking could lead to similar kinematics that were observed for the exercise device, and this should be explored further. However, walking slower does not involve an unstable base of support, or the complex motor control associated with using the exercise device, both of which could be contributing to increased local lumbopelvic muscle recruitment.

CONCLUSION

Key differences between exercising on the device and overground walking included reduced transverse plane range of trunk motion with respect to the pelvis (i.e. increased lumbopelvic stability), a more anteriorly tilted pelvis, and reduced stride length, knee and hip range of motion in the sagittal plane. The greater anterior tilt of the pelvis potentially moved the pelvis into a more advantageous position for the recruitment of TrA and LM. However, the unstable base of support afforded by the new exercise device would seem to add a challenge to movement control that may result in greater TrA and LM activity than overground walking. Future investigations should examine TrA and LM activity during walking and exercising on the new device using ultrasound imaging.

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Table 1. Spatiotemporal characteristics of overground walking and exercise in the standing position on the device. (SD = standard deviation, CI = confidence interval).

Gait Parameter	Overground Walking		Exercise Device		Mean Difference (95% CI)	*P* value
	Mean	±1SD	Mean	±1SD		
Cadence (steps·min^{-1})	110.7	7.2	71.3	2.7	-39.4 (-43.4 to -35.45)	<0.001
Stride Length (m)	1.41	0.09	1.10	0.00	-0.31 (-0.35 to -0.26)	<0.001
Stride Duration (s)	1.09	0.07	1.69	0.06	0.60 (0.55 to 0.65)	<0.001
Speed (m·s^{-1})	1.30	0.13	0.65	0.03	-0.65 (-0.71 to -0.58)	<0.001
Step Width (m)	0.20	0.03	0.23	0.05	0.03 (0.01 to 0.06)	0.018
Stance Phase (%)	59.54	1.66	49.45	2.26	-10.09 (-11.49 to -8.69)	<0.001

Table 2. Angular range of motion of the trunk, pelvis, hip, and knee in all three planes during overground walking and using the exercise device, also including the mean difference between the two conditions. (SD = standard deviation, CI = confidence interval).

Gait Parameter	Walking		Exercise Device		Mean Difference (95% CI)	*P* value
	Mean	±1SD	Mean	±1SD		
Sagittal Plane						
Trunk	3.93	1.80	3.01	1.67	-0.92 (-0.29 to 2.14)	0.126
Pelvis	2.89	0.78	3.69	1.91	0.8 (-1.86 to 0.26)	0.129
Hip	42.54	3.96	33.38	2.28	-9.16 (6.50 to 11.81)	<0.001
Knee	59.88	4.03	45.22	6.02	-14.66 (10.97 to 18.36)	<0.001
Frontal Plane						
Trunk	12.59	3.26	11.21	4.42	-1.39 (-1.07 to 3.84)	0.248
Pelvis	8.29	3.33	8.09	2.70	-0.20 (-1.85 to 2.26)	0.834
Hip	12.67	3.44	8.77	4.64	-3.90 (0.67 to 7.14)	0.021
Knee	16.50	5.91	9.42	5.22	-7.08 (4.93 to 9.22)	<0.001
Transverse Plane						
Trunk	12.55	3.85	3.92	1.14	-8.63 (6.47 to 10.79)	<0.001
Pelvis	12.00	3.28	9.25	4.18	-2.75 (-0.41 to 5.92)	0.083
Hip	16.93	7.34	8.87	2.73	-8.06 (4.95 to 11.17)	<0.001[a]
Knee	20.66	5.37	10.59	3.96	10.07 (7.06 to 13.09)	<0.001

[a] indicates that these data were not normally distributed

Table 3. Mean angular position of the trunk, pelvis, hip and knee in all three planes during overground walking and exercise in the standing position on the device. (SD = standard deviation, CI = confidence interval).

Gait Parameter	Walking		Exercise Device		Mean Difference	
	Mean	±1SD	Mean	±1SD	(±95% CI)	*P* value
Sagittal Plane						
Trunk	-5.37	6.15	-5.43	6.66	0.06 (-3.44 to 3.56)	0.970
Pelvis	9.06	4.06	15.55	6.18	-6.49 (-9.43 to -3.54)	<0.001
Hip	18.30	5.56	40.61	6.62	-22.31(-25.88 to -18.73)	<0.001
Knee	26.28	4.62	50.39	6.69	-24.11 (-26.69 to -21.52)	<0.001
Frontal Plane						
Trunk	-0.41	1.68	0.53	2.05	-0.94 (-2.27 to 0.38)	0.150
Pelvis	-0.25	1.23	-0.33	1.83	0.08 (-0.69 to 0.85)	0.827
Hip	-0.14	2.02	-0.91	2.33	0.77 (-0.13 to 1.67)	0.088
Knee	2.99	3.92	0.44	7.53	2.55 (-0.37 to 5.48)	0.082
Transverse Plane						
Trunk	-2.11	1.86	-1.70	2.03	-0.41 (-1.25 to 0.42)	0.311
Pelvis	-0.65	2.26	-1.63	2.93	0.98 (-0.05 to 2.01)	0.056[a]
Hip	8.77	8.35	2.55	5.59	6.22 (1.77 to 10.66)	0.010[a]
Knee	-8.77	9.14	1.16	8.58	-9.94 (-12.45 to -7.42)	<0.001

[a] indicates that these data were not normally distributed

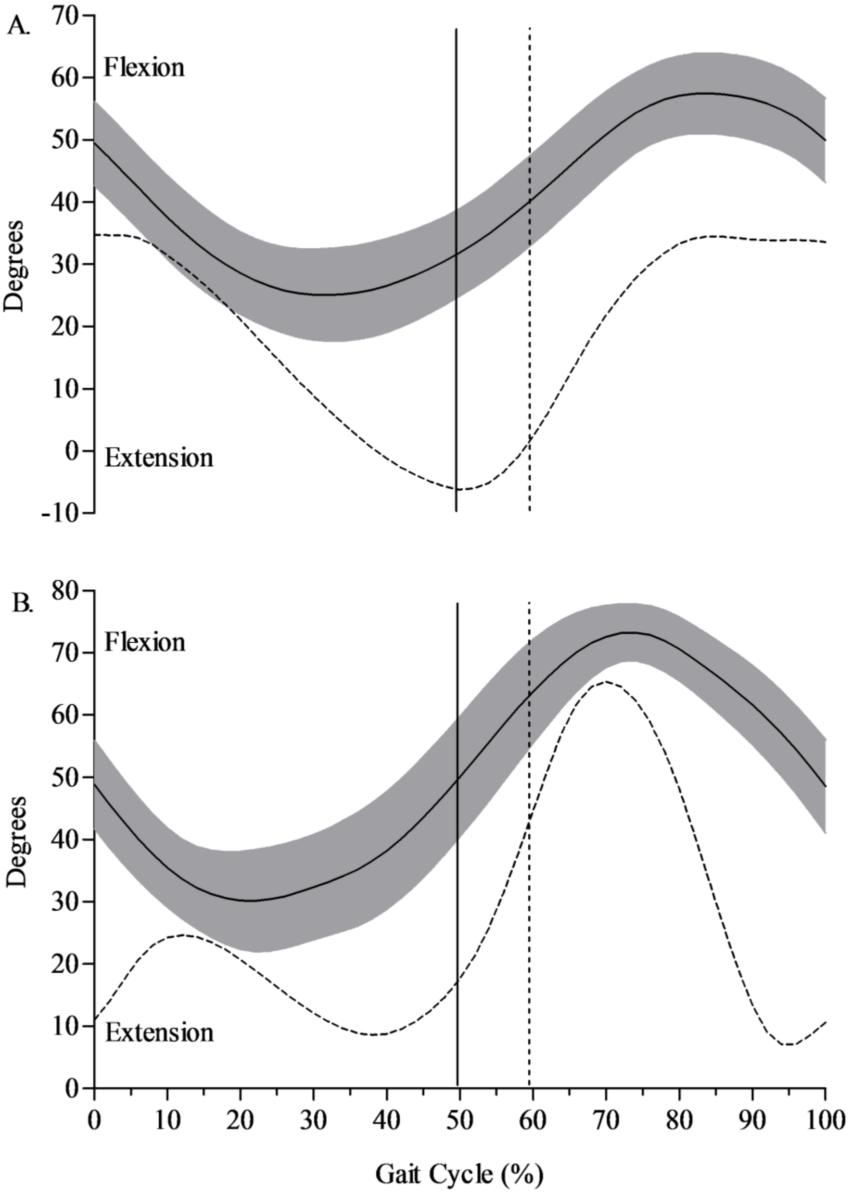
FIGURE CAPTIONS

Figure 1. The exercise device during use.

Figure 2. Hip (A) and knee (B) angles are presented for overground walking (- - -) and exercise (—) conditions. The shaded region represents the standard deviation for the exercise device data series.

Figure 1

Figure 2


A.
70
60
50
40
30
20
10
0
-10
Flexion
Extension
Degrees
B.
80
70
60
50
40
30
20
10
0
Flexion
Extension
Degrees
0
10
20
30
40
50
60
70
80
90
100
Gait Cycle (%)