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A comparison of equine hindlimb muscle activation and joint motion between forward and backward walking

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ABSTRACT

Backward walking (BW) is commonly employed as a physiotherapeutic exercise for horses based on anecdotal evidence for improving hindlimb strength, coordination, and range of motion. However, limited scientific evidence supports these assumed benefits. This study aimed to measure and compare equine hindlimb muscle activity and movement during BW and forward walking (FW). Three-dimensional kinematic and surface electromyography (sEMG) data were synchronously collected from unilateral (left) hindlimb and hip extensor (biceps femoris and gluteus medius) and flexor (tensor fasciae latae) muscles of ten horses during FW and BW. Normalised average rectified value (ARV), peak amplitude (PA), and muscle activity duration were calculated from sEMG data. Spatiotemporal and angular parameters were calculated from kinematic data. Wilcoxon signed rank tests or paired t-tests assessed differences between FW and BW. Compared to FW, significant ($p < 0.001$) increases in hip extensor and decreases in hip flexor muscle ARV and PA were observed during BW. Muscle activity duration was significantly ($p < 0.001$) longer for biceps femoris, and shorter for gluteus medius and tensor fasciae latae during BW. Hindlimb pro/retraction and sagittal plane hip and stifle joint movement cycles were time-reversed between BW and FW. Hindlimb joints were significantly less flexed and extended ($p < 0.001$) during BW, except for peak stifle and hock flexion during swing, and MTPJ extension during stance, which were significantly greater ($p < 0.001$) during BW. Findings support BW as a physiotherapeutic exercise to target increased hip extensor activity during weightbearing and to facilitate increased stifle and hock joint flexion during non-weightbearing phases of the gait cycle.

1. Introduction

Exercise prescription is a fundamental component of human and veterinary physiotherapy practice. The appropriate selection of exercises allows the physiotherapist to create tailored programmes, which address the patient's individual goals. Of these exercises, backward walking (BW) is commonly prescribed by veterinary physiotherapists for equine patients. BW is described as a symmetric, diagonal gait without a suspension phase [1]. In contrast to forward walking (FW), the BW gait cycle is initiated with the limb in a retracted, "toe first" position at hoof impact, with weightbearing and propulsion occurring during the protraction phase of the stride [1]. This reversal of the limb movement cycle and the subsequent caudal shift of the horse's centre of mass during BW is anecdotally believed to activate, and subsequently strengthen, the epaxial, hypaxial and hindlimb muscles, thus improving proprioceptive awareness, strength, and neuromuscular control [1–3]. Despite these assumed benefits, no studies have evaluated the isolated effect of BW on axial or hindlimb movement in conjunction with muscle activity in horses. As such, the efficacy and subsequent selection of this exercise within veterinary physiotherapy is not currently supported by scientific evidence. In recognition of this, we conducted a preliminary study to evaluate the hindlimb movement strategies and underlying

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muscle activation patterns that horses adopt during BW when compared to FW.

Given its comparative infancy, the veterinary physiotherapy sector commonly extrapolates from the human physiotherapy evidence-base to support treatment of animals [4–7]. In humans, scientific evidence supports the use of BW as a physiotherapeutic exercise for improving gait and mobility in patients presenting with a wide range of neurological and musculoskeletal conditions, as well as athletic populations [8–11]. The limb movement patterns during human BW are kinematically similar to FW, but the movements are reversed in time [12–15] and are characteristically initiated by toe, rather than heel strike [16]. Thus, BW has been described as a near temporal reversal of FW [13]. Despite the time-reversed similarities between FW and BW in humans, several kinematic and kinetic differences have been reported. BW gait is performed at a significantly slower speed than FW [12,15,17], but the ratio of stance and swing duration remain consistent across FW and BW gait cycles [12,14,15,18]. In addition, significant reductions in sagittal plane joint range of motion (ROM) and power at the hip, knee and ankle [12,15], and reduced joint loading, particularly at the knee [17,19], have been reported during BW compared to FW in humans.

Electromyographic evaluations of humans reveal that muscle activity is also significantly altered between FW and BW to generate the reversal of the limb movement cycle and to control the movement of the centre of mass [13,14,18,20]. BW elicits significant increases in activity of gluteus medius, rectus femoris, vastus lateralis, biceps femoris and tibialis anterior in humans [13,14,18]. This intensification of limb muscle activation has been linked to the greater metabolic demands of BW [14,21] but has also been explained through the time-reversal of joint powers and surface electromyography (sEMG) patterns, which suggest that the joints and muscles responsible for power generation during FW are responsible for absorption during BW and vice versa [12,13,15]. In other words, the temporal cycling of muscle contractions reverses, and the amplitude of muscle activation changes due to the selective modulation of agonist-antagonist pairing during FW and BW [13]. The switch to an agonistic (concentric) role for the quadriceps and thus decreased force absorption at the knee, as well as increased antagonist (eccentric) action of the hamstrings during BW [13,14] are indicated for patients with knee pain or dysfunction (e.g. anterior cruciate ligament injury and osteoarthritis). Studies have demonstrated improved balance in quadriceps:hamstring strength ratios and knee joint ROM and proprioception in these patient groups when BW is implemented in treatment plans [8,10,22,23]. In addition, BW is commonly used within neurological rehabilitation for patients with Parkinson's Disease or post-stroke [9,24], as the absence of visual feedback has been shown to challenge, and thus improve, balance, coordination, proprioception, and spatial awareness in these patient groups [24,25]. Thus, the clinical benefits related to the temporal reversal of movement and muscle activation patterns that occur during BW have been demonstrated for humans. However, it remains unknown whether these benefits are transferrable to equine patients, particularly those with similar conditions (e.g. osteoarthritis, tendinopathies, or neurological deficits). Furthermore, differences in hindlimb kinetics of quadrupeds compared with bipeds are known to affect muscle function during FW, so muscle activation patterns during BW may be different in horses than in people [26].

In contrast to the human literature, studies evaluating the movement strategies adopted by horses during BW are scarce. A recent study of healthy horses reported a significantly wider base of fore- and hindlimb support, as well as significantly increased thoracolumbar and lumbosacral flexion throughout BW movement compared to FW [27]. These adaptations in axial movement and limb positioning were interpreted as a strategy to improve whole-body stabilisation, particularly as a compensatory method for coping with compromised visual input during BW [27], although horses may be less visually compromised than people due to their larger range of lateral and caudal monocular vision. BW is also used clinically to test for equine shivers and stringhalt and has thus featured in two known studies evaluating the clinical signs of these neurological movement disorders [28,29]. In these experimental studies, the movement patterns of horses with equine shivers and/or stringhalt were compared to those of healthy (control) horses during BW [28,29]. During BW, the control group in the study by Seino [29] exhibited comparable pelvic roll and flexion/extension ROM across all hindlimb joints, except for the hip joint, which was less flexed, when compared to FW. In agreement with Jobst [27], the control horses in Seino [29] also exhibited a slower walking speed and a wider hindlimb stance during BW when compared to FW, but differences between kinematic parameters during FW and BW conditions were not tested for statistical significance within this study group.

Preliminary insight into the kinematic changes that occur during BW in horses have been studied, but further work is required to understand the underlying neuromuscular adaptations that could support the perceived benefits of this exercise for improving muscular strength and coordination. As several studies have successfully used sEMG to non-invasively quantify the activation patterns of superficial hindlimb muscles during FW in horses [30–34], we propose that sEMG could be used to fill this gap in knowledge to determine if, and how, horses alter hindlimb muscle activity during BW, when compared to FW. Thus, this study aims to quantify and compare hindlimb movement and the activity of selected hip flexor and extensor muscles between FW and BW in horses. Based on existing research in the human field, we hypothesise that, in comparison to FW, horses will exhibit a general temporal reversal of sagittal plane hindlimb joint movement during BW, that is perpetuated by increased activation of hip flexor and extensor muscles.

2. Materials and methods

Ethical approval for this study was obtained from the University of Liverpool's Veterinary Research Ethics Committee (Ref. VREC1432) and the University of Central Lancashire's Animal Welfare Ethical Review Board (AWERB) (Ref. Re/23/03). Written informed consent was obtained from the individual with legal authority to provide consent for each horse (i.e., Myerscough College yard manager or horse owner). The study was conducted at Myerscough College (Preston, UK).

2.1. Horses and preparation

Data were collected from a convenience sample of ten horses ($n = 10$) from Myerscough College's institutional herd (mean $\pm$ standard deviation (SD) age: 16.7 ± 3.8 years, height: 156.0 ± 8.4 cm, sex: 4 mares, 6 geldings, breed: various). All horses were in ridden work at the time of the study and had no diagnosed or perceived lameness or neurological disorders/deficits, as disclosed by their owner/rider. Horses underwent three separate training sessions prior to data collection to ensure they were proficient in executing the BW exercise. Horses wore their normal bridle during training and data collection sessions.

sEMG sensor (Trigno, Delsys Inc., USA) sites were prepared so that muscle activation could be recorded from the left gluteus medius (GM), vertebral head of biceps femoris (BF) and the tensor fasciae latae muscle (TFL). Clippers were used to remove hair from sEMG sensor sites which were as follows; GM: approximately halfway between the lumbosacral joint and greater trochanter; BF: approximately over the third trochanter and 9 cm cephalad to the cranial margin of semitendinosus; TFL: approximately one quarter of the distance between the ventral tuber coxae and the lateral epicondyle of the femur [35–37]. Following clipping, all sites were thoroughly cleaned with isopropyl alcohol. A small amount of saline solution was applied to each electrode to act as an electrolytic solution, before sEMG sensors were adhered to the skin using Delsys Adhesive Surface Interface Strips (Delsys Inc., USA) and strips of double-sided carpet tape [38]. Sensors were positioned on the muscle belly, with the electrodes oriented perpendicular to the underlying muscle fibre direction [39,40].

To collect three-dimensional (3D) kinematic data, spherical retroreflective markers (25 mm diameter) were attached over the poll, withers (T5), first lumbar vertebrae (L1), between the tubera sacrale, the most dorsal part of tuber coxae, greater trochanter, lateral epicondyle of the femur, talus, metatarsal epicondyle (centre of rotation of the metatarsophalangeal joint (MTPJ)) and the lateral hoof wall [38,41]. In addition, an inertial measurement unit (IMU) sensor (Trigno, Delsys Inc., USA) was attached to a brushing boot over the left third metatarsal for gait event detection purposes.

2.2. Equipment set up

Eight infrared cameras (Oqus 3, Qualisys AB, Sweden) were positioned side-by-side in a linear configuration and an extended calibration was conducted, which resulted in a calibrated volume of approximately 9 m in length. A runway of approximately 1.8 m width was created using ground poles to define the optimal capture volume and to encourage each horse to move in a straight manner between the poles during BW and FW trials.

2.3. Data collection protocol

Unilateral (left side) sEMG (2000 Hz) and 3D kinematic (200 Hz) data were collected synchronously using Qualisys Track Manager (QTM) software (version 2024.1, Qualisys AB, Sweden) and an external trigger system (Delsys Trigger Module, Delsys Inc., USA). First, a static trial was recorded with each horse standing square in the centre of the runway. Then, FW and BW trials were completed in a randomised order, defined using an online tool [42]. During each trial, horses were led through the runway by the same handler and were permitted to walk at their preferred speed. For each horse, five successful trials were collected per condition (FW, BW). A trial was successful when the horse walked between the placing poles with the body aligned along the direction of movement and at a consistent rhythm.

2.4. Data processing and analysis

Data were tracked in QTM software and imported into Visual 3D (version 2021.06.2, HAS-Motion Inc., USA) software for post-processing of sEMG and kinematic data. Gait events were defined using sagittal plane gyroscopic data from the distal left hindlimb IMU. Left hindlimb impact and lift off events were defined by ascent and descent through a 0°/s threshold, respectively [43]. A stride was defined by consecutive left hindlimb impact events for both FW and BW conditions. For each stride, discrete spatiotemporal variables were calculated, including stride velocity, stride duration and left hindlimb stance duration. Stride velocity was calculated as the first derivative of the tubera sacrale marker coordinates in the direction of movement (laboratory coordinate system, x-axis), which was positive for FW and negative for BW, and averaged over each walk stride.

2.4.1. sEMG data

sEMG signals were DC-offset removed, high-pass filtered using a Butterworth 4th order filter with a 40 Hz cut-off frequency and full-wave rectified [44]. For each muscle, the average rectified value (ARV) was calculated from full-wave-rectified signals using stride duration as the time domain. A root mean square (RMS) filter (window length: 0.125 s, window overlap: 0.121 s) [45] was used to smooth the full-wave rectified signals. Peak amplitude (PA) was extracted from each stride using the RMS filtered signals. Outlier strides were identified and removed by setting outlier limits as ± 2 SD outside of the mean for ARV and PA values within each subject, muscle, and condition [38]. ARV and PA parameters were normalised relative to the maximum associated value across all FW strides within each horse and muscle [45]. This normalisation technique allowed for analysis of the proportional difference in muscle activity between FW and BW.

For each muscle, the timing of sEMG activity onset, offset, and resultant muscle activity duration were calculated. Muscle activity onset and offset events were calculated using the double threshold method; onset and offset were respectively defined as the point where the signal exceeded, or was less than 20 % of the PA of each individual EMG signal, for a time duration greater than 5 % of the average stride duration across all horses [38]. To improve accuracy, and to account for individual variation in baseline activity, the amplitude threshold was increased in 10 % increments for certain horse/condition combinations [38]. Following post-processing, onset and offset events were visually checked by two researchers for accuracy (F.E., L.S.G.) [38]. Muscle activity onset and offset events, and the subsequent activity duration of each muscle, were normalised to stride duration [38].

2.4.2. Kinematic data

Kinematic data were interpolated (maximum gap: 10 frames) and low pass filtered (Butterworth 4th order) with a cut-off frequency of 6 Hz [46]. In accordance with the method described by Hobbs et al. [47], a rigid-body segment model of the left hindlimb was created for each horse using their static trial and was applied to the corresponding dynamic trials. Sagittal plane joint angles were calculated based on the static trial for the hip, stifle, hock, and metatarsophalangeal joints, using the cardan sequence x, y, z. Joint flexion/extension was defined as rotation around the proximal segment coordinate system x-axis, and the flexor side defined as palmar for the stifle and metatarsophalangeal joints and as cranial for the hip and hock joints. Hindlimb pro/retraction angles were calculated using a hindlimb segment in relation to a reference body segment. The reference body segment was defined using markers placed between the tubera sacrale and the withers and the hindlimb segment was defined by markers on the tuber coxae and the lateral hoof wall [48]. Hindlimb protraction and joint flexion angles were defined as positive, with retraction and extension angles defined as negative. Joint angle data were normalised to the corresponding angles from each horse's static trial to correct for conformational differences. Thus, the kinematic data are presented as angular changes from the standing position [48–50]. Peak joint flexion, extension, and pro-retraction angles, as well as the corresponding time of these peaks were extracted from each stride. The time of peak joint angle flexion and extension and hindlimb pro- retraction angles were normalised to stride duration.

2.5. Statistical analysis

Statistical analyses were conducted in SPSS (version 29.0.2.0 (20); IBM, USA). Data were assessed for normality using a Kolmogorov-Smirnov test, which revealed that all sEMG and selected kinematic variables were not normally distributed; therefore, non-parametric tests were used. Descriptive statistics (median ± interquartile range (IQR)) were calculated for non-normally distributed sEMG and kinematic variables data and a Wilcoxon signed rank test was used to establish if there is a significant difference between FW and BW conditions. For normally distributed kinematic variables, mean and SD were calculated, and a paired *t*-test was used to test for significant differences between FW and BW conditions. Significance levels for all tests were set at $p < 0.05$.

3. Results

For sEMG data, a minimum of 102 FW strides and 129 BW strides were analysed for each muscle. For kinematic data, a minimum of 51 FW strides and 87 BW strides were analysed. Data from one horse were excluded from analysis due to its inability to complete a successful BW trial, as defined in 2.3. Therefore, results are presented from nine ($n = 9$) horses.

3.1. sEMG data

Group averaged, enveloped sEMG signals from the studied hip extensor (BF, GM) and flexor (TFL) muscles during FW and BW conditions are illustrated in Fig. 1. Descriptive statistics (median ± IQR) for ARV, PA, and muscle activity duration data across conditions are presented in Table 1. Fig. 2 provides examples of overlaid sEMG signals and angle-time curves from representative horses and strides, thus illustrating the muscle activation patterns in relation to the resultant movement patterns of the selected joints that they work on. sEMG waveforms from individual horses across FW and BW conditions are presented in Supplementary Figures S1 - S3.

3.1.1. Hip extensors: biceps femoris and gluteus medius

BF activity occurred between late-swing and early-stance phase (median onset: 3.5 %, median offset: 17.6 %) (Figs. 1a, 2a) during FW and between early-swing and late-stance phase (median onset: 89 %, median offset: 50.4 %) during BW (Figs. 1d, 2d). The earlier onset and delayed offset of BF during BW resulted in a significant increase in overall muscle activity duration ($p < 0.001$, Table 1), that was nearly three times longer than that observed during FW. These phasic activity patterns for BF were consistent across horses for FW and BW conditions (Figure S1). A single burst of GM activity occurred during FW, between early (median onset: 6.7 %) and late-stance phase (median offset: 48.4 %) (Figs. 1b, 2b). However, during BW a double burst pattern occurred, with the initial burst occurring between mid-stance until late-stance phase (median onset: 29.5 %, median offset: 48.3 %) and the second burst occurring between late stance phase and mid-swing phase (median onset: 69.3 %, median offset: 87.0 %) (Figs. 1e, 2e). Despite the double burst of GM activity during BW, there was a significant reduction in muscle activity duration ($p = 0.002$, Table 1) compared to FW. These phasic activity patterns of the GM were relatively consistent across horses for FW and BW conditions (Figure S2), but this muscle showed the greatest variability in terms of proportional change in sEMG amplitude between conditions, as illustrated by the IQR values in Table 1 and the SD plotted in Figures 1 and S2. Significant increases in ARV and PA were observed for both BF and GM during BW when compared to FW ($p < 0.001$, Table 1).

3.1.2. Hip flexor - tensor fasciae latae

During FW, TFL exhibited an initial short, low amplitude burst of activity during early stance phase (median onset: 10.8 %, median offset: 18.0 %) (Figs. 1c, 2c), but this was intermittent across horses and strides (Figure S3). This was followed by a second, higher-amplitude and consistent, burst of muscle activity from mid-stance to mid-swing phase (median onset: 38.7 %, median offset: 81.8 %) (Figs. 1c, 2c). A similar activation pattern was observed during BW, with a small and intermittent initial burst of activity (median onset: 17.1 %, median offset: 22.4 %), followed by a consistent, higher amplitude burst of muscle activity that occurred comparatively later from late-stance to late-swing phase (median onset: 61.2 %, median offset: 83.8 %) (Fig. 1 f, 2 f, S3). The main burst of TFL activity was characterised by relatively higher amplitude during stance phase, and lower amplitude during early swing phase in FW, but this pattern was reversed in BW (Figs. 1c, f, 2c, f). During BW, the ARV, PA, and muscle activity duration of TFL were

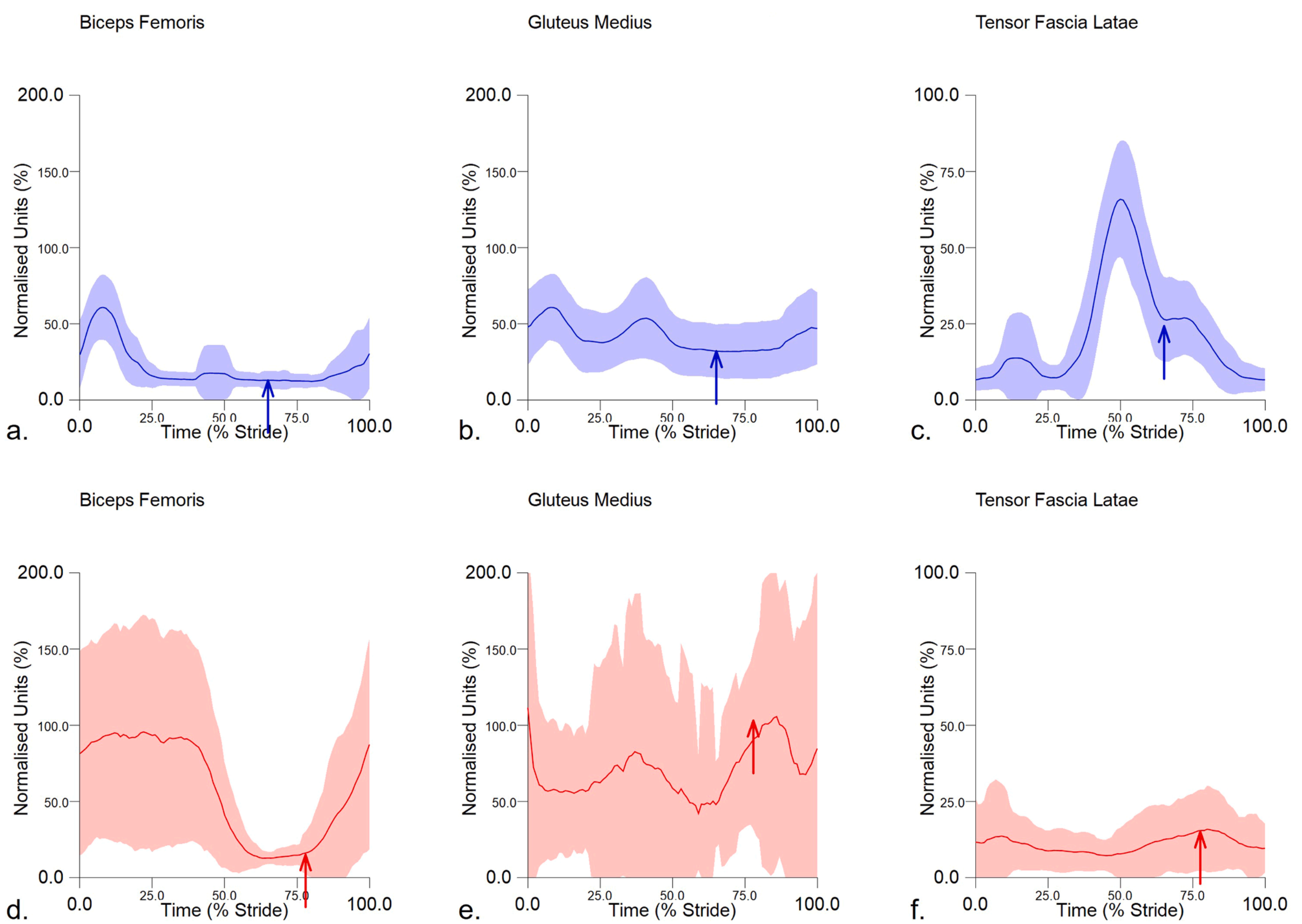

Fig. 1. Mean (solid line) and standard deviation (shaded area) time and amplitude-normalised, linear enveloped sEMG signals from (a, d) biceps femoris, (b, e) gluteus medius, and (c, f) tensor fasciae latae during (a, b, c) forward walking (blue) and (d, e, f) backward walking (red) conditions. sEMG signals are DC-offset removed, high-pass filtered (40 Hz cut-off) and smoothed using an RMS filter (window length: 0.125 s, window overlap: 0.121s). Upward arrows represent the average left hindlimb lift-off event. Data are time-normalised to stride duration, calculated using corresponding impacts of the left hindlimb.

Table 1
Descriptive statistics (median and interquartile range (IQR)) for average rectified value (ARV) peak amplitude (PA), both presented as normalised units (%), and muscle activity duration (% stride duration) across forward and backward walking conditions for (n = 9) horses. Between-condition significance is presented for each variable with p-value.

Muscle	Parameter	Forward Walking		Backward Walking		*p*-value
		Median	IQR	Median	IQR	
Biceps Femoris	ARV (%)	79.5	69.8–93.4	187.1	131.9–298.0	< *0.001*
	PA (%)	68.5	52.8–85.8	114.2	76.7–187.1	< *0.001*
	Activity Duration (%)	19.6	14.3–24.8	58.6	50.9–65.5	< *0.001*
Gluteus Medius	ARV (%)	89.3	81.7–94.8	135.1	108.2–166.2	< *0.001*
	PA (%)	83.9	66.3–94.5	132.5	101.1–207.1	< *0.001*
	Activity Duration (%)	47.5	42.6–54.2	34.6	21.9–53.1	= *0.002*
Tensor Fasciae Latae	ARV (%)	78.4	68.8–85.9	31.4	25.2–38.9	< *0.001*
	PA (%)	78.3	64.8–87.2	21.8	14.7–33.6	< *0.001*
	Activity Duration (%)	43.2	40.2–54.1	34.3	21.3–51.8	< *0.001*

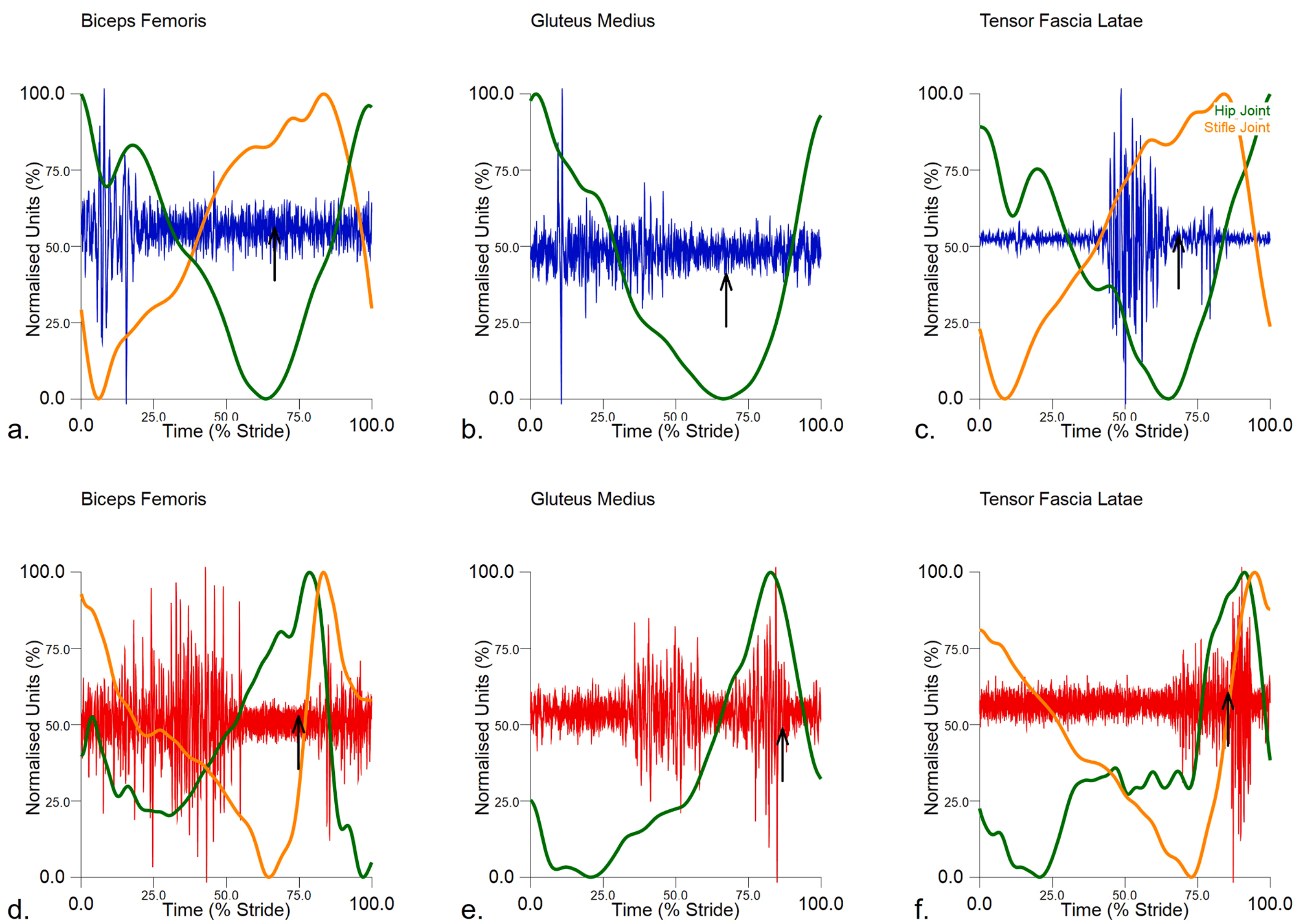

Fig. 2. Phasic activity patterns of (a, d) biceps femoris (Horse 7), (b, e) gluteus medius (Horse 3), and (c, f) tensor fasciae latae (Horse 5) muscles and normalised sagittal plane angles, for the joints (hip, stifle) that each muscle works on, from representative horses during forward (blue) and backward (red) walk strides. Green and orange solid lines represent hip and stifle joint angle data, respectively. Data are presented as normalised units (%) to permit overlay of sEMG and kinematic signals, so amplitude comparisons of sEMG data cannot be made. sEMG signals are direct current (DC)-offset removed and high-pass filtered (Butterworth 4th order, 40 Hz cut-off frequency). Kinematic signals are low-pass filtered (Butterworth 4th order, 6 Hz cut-off frequency). Upwards arrows indicate the hoof lift-off event. Data are time-normalised to stride duration, calculated using corresponding impacts of the left hindlimb.

significantly decreased ($p < 0.001$, Table 1) compared with FW conditions.

3.2. Kinematic data

Descriptive statistics (mean ± SD, or median ± IQR) for kinematic data across FW and BW conditions are presented in Table 2 and group averaged, angle-time curves are illustrated in Fig. 3. During BW, stride velocity significantly decreased, and stance duration and duty factor were significantly greater compared with FW ($p < 0.001$, Table 2). The hindlimb impact event occurred when the limb was in a protracted position during FW and in a retracted position during BW, which coincided with the subsequent temporal reversal of proximal joint (hip and stifle) movement in the sagittal plane (Fig. 3). In contrast to FW, the

Table 2
Descriptive statistics (mean ± standard deviation (SD) or median ± interquartile range (IQR)) for discrete kinematic variables for forward walking and backward walking across (n = 9) horses . Between-condition significance is presented for each variable with p-value.

Parameter	Unit	Forward Walking		Backward Walking		*p*-value
		Median	IQR	Median	IQR	
Stride Velocity	m/s	1.5	1.4–1.6	−0.5	0.6–0.4	*< 0.001*
Stride Duration	s	1.3	1.2–1.3	2.0	1.7–2.3	*< 0.001*
Stance Duration	s	0.8	0.8–0.9	1.5	1.3–1.8	*< 0.001*
Duty Factor	%	65.9	64.5–67.4	78.6	72.5–83.8	*< 0.001*
Hindlimb Protraction	°	16.9	14.2–25.6	13.4	10.5–24.2	*< 0.001*
	% Stride	4.0	2.8–4.7	69.8	66.7–73.8	*< 0.001*
Hindlimb Retraction	°	−24.5	−28.6 - −17.1	−8.3	−12.9 - −2.9	*< 0.001*
	% Stride	68.0	66.1–69.6	97.6	93.3–98.9	*< 0.001*
Hip Flexion	°	12.8	11.1–15.1	8.3	5.2–11.7	*< 0.001*
	% Stride	99.2	98.0–100.4	78.4	73.4–84.6	*< 0.001*
Hip Extension	°	−5.2	−6.9 - −2.6	−3.8	−6.0 - −0.7	*< 0.001*
	% Stride	62.7	60.6–64.8	12.6	8.8–19.7	*< 0.001*
Stifle Flexion	°	21.1	13.9–24.9	31.2	26.8–38.7	*< 0.001*
	% Stride	83.3	82.5–84.5	91.0	87.0–94.1	*< 0.001*
Stifle Extension	°	−20.5	−26.7 - −13.2	−10.4	−22.8 - −5.9	*< 0.001*
	% Stride	6.5	5.9–7.1	67.2	63.6–71.7	*< 0.001*
Hock Flexion	°	30.1	27.4–32.2	36.7	33.5–43.3	*< 0.001*
	% Stride	90.1	89.0–91.0	88.7	84.6–91.1	*0.03*
Variable		Mean	±SD	Mean	±SD	*p*-value
Hock Extension	°	−7.5	1.7	−3.0	2.8	*< 0.001*
	% Stride	58.6	4.1	65.7	5.8	*< 0.001*
MTPJ Flexion	°	49.2	6.6	49.3	7.1	*0.917*
	% Stride	79.8	5.5	86.0	4.5	*< 0.001*
MTPJ Extension	°	−11.0	2.2	−13.0	3.0	*< 0.001*
	% Stride	52.1	3.3	20.8	7.9	*< 0.001*

hindlimb underwent protraction during the significantly prolonged BW stance phase, that culminated with significantly less peak protraction, hip flexion and stifle extension angles, than those which occurred at the beginning of FW stance ($p < 0.001$, Table 2). Similarly, the hindlimb was retracted during the shortened BW swing phase, that culminated with significantly less peak retraction and hip extension angles, than those which occurred at the beginning of FW swing ($p < 0.001$, Table 2). Flexion of the stifle joint occurred during swing phase in both FW and BW, but significantly greater and delayed peak stifle flexion was observed during BW ($p < 0.001$, Table 2, Fig. 3c).

In contrast to the proximal limb joints, the flexion/extension pattern of the hock and metatarsalphalangeal joints were not time reversed during BW (Fig. 3d, e). During the prolonged BW stance phase, peak MTPJ extension was significantly greater and occurred significantly earlier than in FW, and vice versa for peak hock joint extension ($p < 0.001$, Table 2). During the shortened BW swing phase, peak hock flexion significantly increased ($p < 0.05$, Table 2) and occurred significantly earlier ($p < 0.001$, Table 2), whereas MTPJ flexion occurred significantly later, but the flexion angle did not significantly differ ($p > 0.05$) from FW.

4. Discussion

Despite its common use as a physiotherapeutic exercise for horses and its assumed benefits for improving hindlimb strength and ROM, the effect of BW on hindlimb muscle activation and movement in horses had not been quantified. As such, this represents the first study to combine sEMG and kinematic data to compare the movement strategies that horses adopt during BW. We observed significant increases in the activity of the selected hip extensors (BF and GM) and significant decreases in the hip flexor (TFL) muscles studied here during BW in comparison to FW. There was a general temporal reversal of hindlimb pro- retraction and proximal joint (hip and stifle) flexion/extension movement patterns during BW, but this was not observed in the distal limb joints (hock and MTPJ) that exhibited similar movement patterns to FW. In addition, sagittal plane ROM generally decreased during BW across all hindlimb joints, except for peak stifle and hock flexion and MTPJ extension angles, which were significantly greater during BW. Consequently, our hypothesis can be partially accepted, as a temporal reversal of proximal hindlimb joint movement was observed, which was perpetuated by significant increases in BF and GM, but not TFL, muscle activity.

Our electromyographic and kinematic findings for FW were largely in agreement with existing equine biomechanics studies. The BF and GM are active from late-swing phase to mid-late stance phase and act to extend the hip joint, and in the case of the BF to stabilise the stifle joint, which was observed here (Fig. 2a, b) and in previous sEMG and kinematic studies of horses during FW [32–34,51]. Conversely, the TFL acts as a hip flexor and stifle extensor and its activation has been described as contributing to the initiation of swing phase and limb protraction [26, 34]. The sEMG data from this study, and others [32], support this function through a main burst of TFL activation occurring from mid-stance to mid-swing phase [32,51]. In addition, our findings coincide with those of Tokuriki and Aoki [32], who observed higher sEMG amplitude during late stance phase, and comparatively lower amplitude during early swing phase. As such, our sEMG and kinematic data support the notion that the TFL may not function primarily to flex the hip joint during swing phase, but rather to stabilise the stifle joint as it flexes during late-stance phase, through tensioning the fascia latae that surrounds the joint [32]. Thus, although we have grouped and generally refer to the muscles studied here as either hip flexors or extensors, it is important to note that they are not limited to working on sagittal plane movement of the hip joint and our findings support their other functional roles. Taken together, our findings agree with previous reports of hindlimb muscle activity and resultant movement patterns during FW, which supports the validity of our data.

No known studies have quantified muscle activity during BW in horses. Further, only one known study has reported hindlimb kinematic data in a small control group of healthy horses during BW [29], which makes comparisons of our findings with other work difficult. The horses in this study adopted a significantly slower BW gait, compared to FW, which agrees with other equine [29] and human studies [12,15,17]. Although not tested for statistical significance, Seino [29] reported that peak hindlimb joint flexion/extension angles and ROM of healthy control horses were comparable during BW and FW, except for the hip joint that showed decreased ROM during BW [29]. This agrees with our observation that the hip joint was significantly less flexed and extended

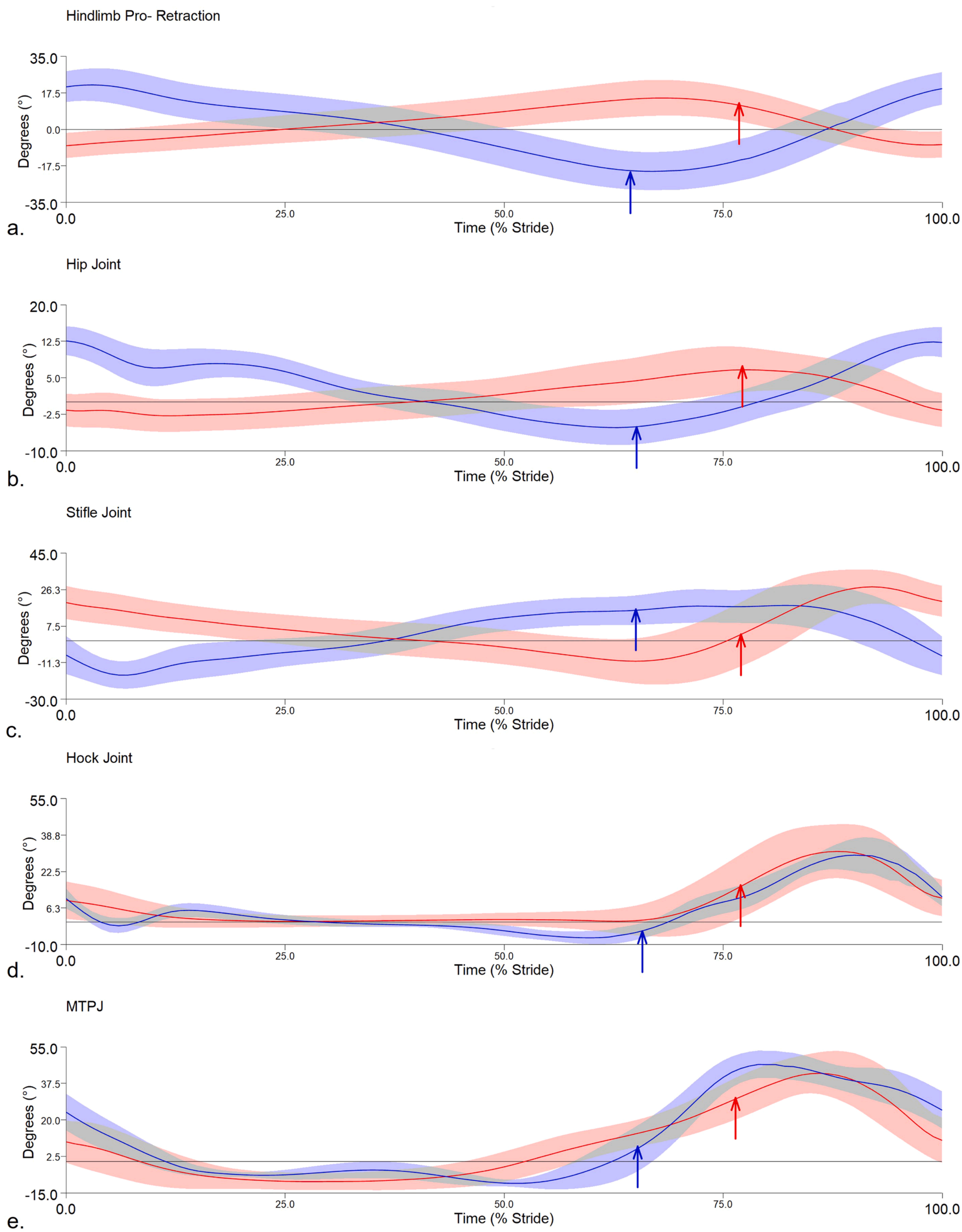

Fig. 3. Mean (solid line) and standard deviation (shaded area) for normalised sagittal plane joint- time-angle curves for (a) hindlimb pro/retraction, (b) hip, (c) stifle, (d) hock and (e) metatarsophalangeal (MTPJ) joints during forward (blue) and backward (red) walking conditions. Signals are low-pass filtered (Butterworth 4th order, 6 Hz cut-off frequency). Positive values indicate flexion/protraction and negative values indicate extension/retraction. Upward arrows represent the average left hindlimb lift-off event. Data are time-normalised to stride duration, calculated using corresponding impacts of the left hindlimb.

during BW. In addition, the descriptive statistics from Seino [29] show a general trend for less peak flexion and extension across all hindlimb joints in their healthy control group, except for stifle and hock, and MTPJ, which were more flexed and extended, respectively, and agrees with our findings. Our kinematic findings also agree with those described in humans during BW, namely decreased joint ROM during BW for the hip and knee joints and a temporal reversal of the flexion/extension cycle when compared to FW [12–14]. In contrast to humans, the angle-time curves for FW hindlimb pro-retraction and hip and stifle were not mirror images of the time-reversed BW curves, as presented for comparative purposes in Supplementary Figure S4. This is likely related to the fact that humans maintain a similar stance:swing phase ratio during FW and BW [12,14,15,18], whereas horses employ a significantly prolonged stance and shortened swing phase during BW, as observed here and in other studies of healthy horses [29], which would preclude mirror image angle-time patterns across FW and BW in horses.

Our data show, for the first time, that horses significantly modify hindlimb muscle activation to facilitate the kinematic changes, namely the reversal of proximal limb motion, that were observed here during BW. Like humans [13,14,18], horses exhibit increased activation of the hip extensor muscles during BW, as quantified by significant increases in sEMG amplitude parameters from the BF and GM muscles. BF was active during early stance phase in both FW and BW but remained active for significantly longer during BW until approximately mid-stance, which coincided with the first burst of GM activation to facilitate hip flexion, stifle extension, and subsequent limb protraction [1]. Interestingly, co-activation of GM and BF was observed during early-stance in FW and in mid-stance in BW, which supports the suggestion that the same muscles are responsible for stabilising the limb and preventing abduction, as it undergoes hip flexion and stifle extension during limb loading and stabilising the trunk as it moves over the limb in both forward and backward gaits [1]. The TFL was active during late-stance and continued through to early swing phase to facilitate hip flexion during FW and BW, but the amplitude of this activation burst was significantly lower during BW. This discrepancy is likely related to the hip being significantly less flexed during BW, as well as co-activation with GM during late stance and with BF during swing phase, which was not observed during FW and may mitigate active contributions from the TFL. Activation of the hip extensors during early-mid swing phase in BW was not observed during FW and likely reflects active muscular contractions from the BF and GM to rapidly extend the hip and flex the stifle joint in preparation for hoof impact. Indeed, the stifle was significantly more flexed during BW, probably to compensate for significant decreases in hip flexion, and coincided with earlier onset of the BF during swing phase supporting its role as a stifle flexor [51].

Human studies have reported a general reversal of concentric/eccentric or agonist/antagonist muscle activity in hip flexor and extensor muscles between FW and BW. Based on our data, it is not possible to conclusively define contraction type as the relationship between sEMG amplitude and muscular force/contraction type becomes distorted during the dynamic nature of the tasks performed here [52]. However, the time synchronised sEMG and kinematic data that are presented here allow us to make careful inferences about the relationship between joint movement and muscle activation, which suggest that the hip extensors do undergo periods of both eccentric and concentric activity during BW. In agreement with Denoix [1], our data suggest that the BF and GM act eccentrically during BW stance phase as a result of increased lumbosacral flexion [1,27] and to aid the trunk moving caudally over the hindlimb through stabilisation of the hip and stifle joints as they undergo flexion and extension, respectively. Concentric activation of the BF and GM during late stance and swing phase of BW have been described to initiate hip extension and subsequent limb retraction during BW [1], which our findings also agree with. This is in contrast to FW, where limb protraction is initiated at the end of stance by contraction of hip flexors, and is maintained by passive recoil of the hindlimb throughout swing phase [46]. Thus, our findings suggest that active contribution from the TFL during late-stance and swing phase is reduced during BW and is compensated for by active muscular contraction of the hip extensors to generate power to initiate swing phase and retraction of the limb. This agrees with human BW where the main propulsive force during BW is provided by the hip extensor muscles [14]. However, further research is required to confirm this theory in horses through the evaluation of joint moments and powers during BW.

An interesting finding that contrasts with that of human BW research, is the non-reversal of distal limb joint angle-time curves between BW and FW in horses. This is relatively unsurprising considering the anatomical differences between species, namely the largely passive role of the suspensory apparatus that supports the distal limb joints in horses [26]. In agreement with Hodson et al. [46], peak MTPJ extension occurred during the latter half of FW stance phase, but in BW, the MTPJ was significantly more extended during the early part of stance. In addition, the characteristic MTPJ double extension pattern that reflects the vertical loading pattern of the limb during FW [46,53,54] was less pronounced, or absent, across horses during BW. There are several potential explanations for these discrepancies in MTPJ movement between FW and BW. Firstly, FW is a 4-beat gait and the MTPJ extension peaks coincide with periods of bipedal support, separated by a period of flexion (dip) during tripedal support, which is not present during 2-beat, diagonal BW gait [1,27]. Secondly, the dip between vertical ground reaction force (GRF) peaks is affected by velocity, nearly disappearing at very slow FW speeds comparable to those observed here for BW [55]. Interestingly, vertical GRF data from humans during FW and BW are also characterised by two peaks, which are nearly symmetrical in amplitude during FW, but the first and second peaks show comparatively higher and lower amplitude, respectively, during BW [12,14,15,17,56]. The higher initial vertical GRF peak in humans is related to the requirement for supporting whole-body weight during limb loading [15] with the smaller second peak relating to the requirement for less propulsive force during late stance [56]. If MTPJ extension reflects the loading pattern of the limb, then our findings suggest that, in accordance with humans, the significantly slower, diagonal BW gait elicits increased vertical loading during early stance, followed by gradually less loading over the significantly prolonged stance phase when compared to FW. However, further work is required to confirm this interpretation using GRF data.

This preliminary study is not without limitations that should be considered when interpreting its findings. We employed a small sample of healthy horses, which we considered to be sufficient for preliminary research. However, future research should build on this work by employing a larger sample, particularly given the greater variability observed in kinematic and sEMG data during BW in horses, which are also features of BW in people [13,15]. Future research is also required to evaluate whether our findings from healthy horses are transferrable to horses presenting with musculoskeletal and neurological conditions that are believed to benefit from BW exercise. We did not control for walking speed in this study, which can be considered a limitation as this is known to influence kinematic and sEMG parameters [57]. However, standardisation of walking speed could only be accomplished by using a treadmill and we did not deem this as appropriate or safe for the BW trials. As such, we attempted to mitigate variation in walking speed by using the same handler throughout all trials and our stride velocity data are indeed in accordance with those reported in previous overground FW [58] and BW [29] studies in healthy horses, which supports the external validity of our findings within an equine rehabilitation setting. In addition, our study was limited to the unilateral evaluation of three superficial hindlimb muscles. Thus, we suggest that future work quantifies bilateral movement and the activation of additional hindlimb muscle groups, like the quadriceps, which demonstrate adaptive activation patterns during BW in humans [14]. Finally, it would be prudent for further research to evaluate the clinical efficacy of BW in regard to the optimal frequency for achieving specific rehabilitation goals (e.g. improved muscle strength) and the effects of BW in comparison with other exercises that are known to increase hindlimb muscle activity, such as gradient walking [30,31].

5. Conclusion

This is the first study to evaluate hindlimb muscle activity and movement during BW in horses. The results demonstrate that BW is beneficial to target increased activation of the BF and GM to stabilise the limb during weightbearing, and to facilitate limb retraction and increased stifle flexion during non-weightbearing phases of the stride cycle. Our findings also suggest that BW could be applied clinically as an active range of motion exercise, to increase joint flexion and range of motion in the non-weightbearing stifle and hock joints. In addition, our findings suggest that BW does not represent a near temporal-reversal of FW as described in humans [12–15] which highlights the importance of species-specific research to inform veterinary physiotherapy practice. This preliminary research provides novel, quantitative data that offers evidence for altered movement and neuromuscular control during BW and suggests it has value within equine physiotherapy and sport conditioning programmes. This study provides an important contribution to the ongoing development of evidence-based practice for veterinary physiotherapists, and it is hoped that future work will build upon this initial study to develop a more comprehensive understanding of whole-body mechanics during BW.

CRediT authorship contribution statement

Fleur Eldridge: Writing – review & editing, Writing – original draft, Visualization, Validation, Software, Resources, Project administration, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Lindsay St. George:** Writing – review & editing, Writing – original draft, Visualization, Validation, Supervision, Software, Resources, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Melanie Chapman:** Writing – review & editing, Supervision, Resources, Project administration, Conceptualization. **Lynne Harrison:** Writing – review & editing, Validation, Conceptualization. **Gillian Tabor:** Writing – review & editing, Validation, Conceptualization. **Charlotte Uttley:** Writing – review & editing, Resources, Conceptualization. **Hilary Clayton:** Writing – review & editing, Validation, Supervision, Conceptualization.

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Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.eqre.2025.100036.

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