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The effect of altered stride length on iliocapsularis and pericapsular muscles of the anterior hip: An electromyography investigation during asymptomatic gait

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ABSTRACT

Background: Anterior pericapsular muscles potentially contribute to active hip joint stability in walking by controlling anterior femoral translation in peak extension. Alternatively, these muscles could flex the hip and tension the anterior capsule to aid initiation of swing. Although never investigated, the location of Iliocapsularis and its extensive anterior hip capsule attachment imply a potential role in these functions. We hypothesised if these muscles contribute to control of femoral head translation (rather than hip flexion), their activation would decrease when peak hip extension is reduced during shorter stride walking.
Research questions: To provide the first description of iliocapsularis activity during walking and challenge the hypothesised roles of the anterior pericapsular hip muscles in gait by walking with shortened strides.
Methods: Fifteen healthy volunteers walked with self-selected and shortened stride lengths. Electromyographic recordings were made from iliocapsularis, iliacus and anterior gluteus minimus with intramuscular electrodes, and rectus femoris with surface electrodes. Stride characteristics were measured using force sensors and 3D motion capture. Ensembles of burst activity profiles for each stride length were compared.
Results: Iliocapsularis displayed a consistent EMG burst around toe-off (terminal hip extension) that peaked during early swing phase with self-selected strides. In shortened strides, iliocapsularis EMG increased during mid to late stance ($p = 0.03$), with no difference in other muscles. Iliocapsularis, iliacus and rectus femoris activity decreased during early stance ($p < 0.01$) in short strides, whereas gluteus minimus EMG increased ($p = 0.03$).
Significance: Iliocapsularis displayed an EMG burst around toe-off during walking, and greater EMG during mid-late stance in short stride walking, which was not seen in other pericapsular muscles. Shortened strides increase the demand for active tensioning of the hip joint capsule in initial swing, and suggests a role for iliocapsularis during active hip flexion in pain free young individuals.

1. Introduction

Anterior pericapsular muscles have the potential to contribute to active hip joint stability in walking by supporting the hip capsule and controlling femoral head translation in peak hip extension. Three muscles that cross the hip have direct attachments to the anterior hip joint capsule; Iliocapsularis, anterior gluteus minimus and rectus femoris. Together with the Iliofemoral ligament these muscles are thought to form a “stability arc” to constrain anterior translation of the femoral head [1]. Iliacus, passes immediately anterior to the joint and is also proposed to restrain anterior translation in hip extension [2,3]. Despite proposed roles in hip joint stability, there has been limited investigation of anterior pericapsular hip muscles during dynamic tasks such as walking.

Electromyography (EMG) of anterior hip muscles has been investigated in gait [4,5], but with a principle focus on understanding their contribution to hip flexion torque [5,6]. Hip flexion is primarily initiated by stored elastic energy from anterior hip structures [7], with activation of rectus femoris and iliacus serving to sustain hip flexion during swing. Iliacus is proposed to have a role in hip joint stability by restraint of the femoral head through its muscle belly, when moving into hip extension [2,3,5]. Similarly, anterior gluteus minimus muscle

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activity in mid to late stance is thought to stabilise the head of femur and counteract anteriorly directed joint forces during gait [4]. Iliocapsularis action in gait is currently unknown. It is proposed to support the Iliofemoral ligament [1] when taught during peak extension, combining to resist anterior femoral head translation [8] in terminal stance. Iliocapsularis alignment with the vertical limb of the Iliofemoral ligament [1] supports this hypothesis but its large direct capsular attachments [9] may also serve to dynamically tension the anterior capsule [10]. Current recordings of Iliocapsularis EMG are limited to isometric hip torque across three planes of movement [11]. The greatest amplitude of activity was recorded during maximum resisted isometric hip flexion. Iliocapsularis' location anterior to the centre of rotation of the hip joint, implies a role in the generation of hip flexion torque. Yet its small cross-sectional area, limited moment arm and attachments close to the joint line suggest it may not be efficient in this role.

Whether anterior pericapsular muscles, including iliocapsularis, act to stabilise the hip, or aid in hip flexion through direct action, or via tensioning the capsule to store elastic energy, requires resolution through direct measurement of muscle activity during gait. When the hip is in terminal extension, during late stance phase [12], the anterior capsule is under its greatest tensile load [1] and the femoral head is exposed to anteriorly directed joint force relative to the acetabulum [6]. Elevated EMG of the anterior pericapsular muscles (including iliocapsularis) at this point in the gait cycle would support a hypothesised role in joint stability during natural gait. Further, if this is the primary function of these muscles, reduced EMG would be expected if hip extension is decreased by walking with shortened stride length [12].

The primary aim of this study was to provide the first description of iliocapsularis EMG during walking, to investigate its pattern of activity and identify whether this supports a hypothesised role in dynamic stability of the hip. The second aim was to compare EMG of all anterior pericapsular hip muscles (iliacus, iliocapsularis, rectus femoris and anterior gluteus minimus) between natural self-selected and shortened strides. It was hypothesised that a role for the pericapsular muscles in augmentation of anterior stability during walking would be supported if burst activity was reduced as the hip approached terminal stance in shortened strides relative to natural self-selected strides.

2. Methods

2.1. Participants

Fifteen healthy and physically active volunteers (10 men, mean (SD) age - 22 (2) years), with no history of hip pain, surgery or pathology, and no lower limb injuries requiring treatment in the preceding 3 months were recruited. The current level of sporting activity for each participant was ascertained using the hip sport activity scale [13]. Demographic data are presented in Table 1. Participants were drawn from the same sample as a previous study of maximal voluntary isometric contractions [11]. All participants provided informed consent, and the institutional Medical Research Ethics Committee (2013001448) approved the study.

Table 1
Participant Demographics, mean (standard deviation).

Sex	Number	Age (yrs.)	Height (cm)	Mass (kg)	BMI (kg/m^2)	HSAS
Male	10	22.4 (1.8)	181.5 (6.0)	79.8 (9.8)	24.3 (3.8)	4.9 (1.6)
Female	4	22.5 (2.4)	171.3 (5.2)	64.3 (7.0)	21.9 (1.6)	6.5 (1.0)
Total	14	22.4 (1.9)	178.5 (7.4)	75.4 (11.4)	23.6 (3.4)	5.4 (1.6)

Abbreviations: BMI body mass index; cm centimetres; HSAS Hip sports activity score (range from 0 = no recreational or competitive sports to 8 = elite competitive sports); kg kilograms; kg/m^2 kg per metre squared; yrs years.

2.2. Electromyography

Three bipolar intramuscular fine-wire EMG electrodes were fabricated by passing 2 strands of Teflon coated stainless steel wires (length - 25 cm) through a 7 cm hypodermic needle (23 gauge, Terumo Medical). After sterilisation, electrodes were inserted into anterior gluteus minimus, Iliocapsularis and iliacus of the "stance dominant" leg. Stance dominance was determined through a previously established protocol [14]. The insertion procedure for Iliocapsularis and anterior gluteus minimus have been previously described in detail [11,15]. Iliacus insertion was developed with reference to published intramuscular insertions [7,16] and cadaveric observations. In brief, iliocapsularis and iliacus insertions were performed in supine and anterior gluteus minimus in side lying. Insertion sites were marked based on anatomical landmarks and electrode insertions guided by real-time B-mode ultrasound (7–10 MHz a linear transducer; LogIQV2, GE Medical Systems Co. Ltd, China), to confirm electrode placement within the target muscles while avoiding local neurovascular structures. After needle removal, the wires were secured to the skin and connected to a Trigno™ Wireless EMG System (Delsys Inc., Boston, USA). Surface electrodes to record rectus femoris EMG were placed according to SENIAM guidelines [17]. The quality of recordings was assessed prior to testing through observation of activation during gentle isometric contractions of the target muscle. EMG signals were recorded using the Trigno™ Wireless EMG System (CMRR > 80 dB@60 Hz; gain – 1000x; bandpass filter - 20–900 Hz), with data sampled at 2000 Hz using Vicon Nexus 1.8.5 (VICON, Oxford Metrics, Oxford UK).

2.3. Measurement of gait parameters

Temporal stride characteristics used for EMG analysis, including time of heel strike and toe-off, were determined by force sensitive resistors (Delsys Inc., Boston, USA), located on the heel and toe of each foot. Retroreflective markers were placed on the posterior of the calcaneus (heel), and first sacral vertebrae (sacrum), to measure, stride length and walking velocity through an eight-camera 3D motion capture system (VICON, Oxford Metrics, Oxford, UK) [18].

2.4. Experimental protocol

Walking trials were recorded within a motion capture volume, with additional space provided before entering and on exiting the recording area to ensure participants travelled at a constant speed. Participants walked continuously for 3 min prior to testing to become accustomed to the intramuscular electrodes and the walking paradigm, and to confirm the quality of EMG signals. Four trials were conducted with participants walking at their self-selected natural stride. Mean stride length was determined by the distance between heel markers in consecutive strides, and mean velocity was determined by the speed of the sacral marker through the motion capture space, across four trials. A stride length of 90% of the participants self-selected stride was determined, and participants were instructed to "walk with smaller steps" for a further four trials. Feedback was provided by the investigator and trials repeated if the stride length was greater than 90% of the mean length during self-selected walking or if gait velocity varied by more than ± 5% between repetitions.

2.5. EMG processing

EMG data were exported and processed using R statistical software (Version 3.3.2 http://www.R-project.org/). DC offset was removed; then EMG data were high-pass filtered (4th order Butterworth, no phase lag; 20 Hz cut-off for surface electrodes, and 50 Hz cut-off for fine-wire electrodes), rectified, and smoothed using a 20-ms moving window. EMG data were time normalised to 101 points (% of the gait cycle) and averaged over two consecutive strides across four trials (eight strides in

total) during natural self-selected gait to create an ensemble average for each participant. Individual participant's ensembles were normalised to peak EMG activity within natural self-selected walking and then averaged across participants to generate an overall EMG profile for iliacus, iliocapsularis, anterior gluteus minimus and rectus femoris [19]. This process was repeated in trials with shortened strides, with individual participant ensembles normalised to peak activity during natural self-selected walking.

The overall profile was then analysed based on previously published phases of activity during the gait cycle. Each stride was separated into early stance phase (0–30%), late stance phase (30–60%), initial to mid swing (60–90%), total stance phase (heel strike to toe-off) and total swing phase (toe-off to subsequent, ipsilateral heel strike). Intra-participant variability in muscle activity during gait was calculated using the mean coefficient of variation (CoV) within each participant across eight strides [14,20]. The CoV was determined for each one percent of the gait cycle for each participant, before being averaged within a defined phase; 0–30%, 30–60%, 60–90%, swing and stance.

2.6. Statistical analysis

Normality of the data was determined by the Shapiro-Wilks test and observations of histograms and Q-Q plots. Gait characteristics were compared between short and long strides using paired t-tests. The EMG data were not normally distributed and consequently analysed using non-parametric statistics. To address the primary aim, the profile of iliocapsularis EMG was inspected to identify the pattern of burst activity in relation to specific phases of the gait cycle. To address the secondary aim, EMG of each muscle was compared between walking conditions (natural vs. short strides) using the Wilcoxon sign rank test for repeated measures. Effects size estimates were calculated to indicate the magnitude of effect by dividing the *z* value, by the square root of the sample size, where 0.2, 0.5 and 0.8 indicate small, moderate and large effects sizes respectively [21]. All statistical comparisons were completed using the Stata statistical software package (Release 13 (64-bit). College Station, TX: Statacorp LP.), with an alpha level of 0.05.

3. Results

Data from one participant and data for a single muscle (Iliocapsularis) from another participant were excluded because of artefact (n = 14 (10 male)). Hip sports activity score was 5.4 (1.6), representing recreational (to competitive) level athletes [13]. Gait velocity was faster for self-selected walking (1.2 (0.1) m/s) than walking with shortened stride (1.0 (0.1) m/s; $p < 0.01$). Stance time increased from self-selected strides (66.7 (1.4)%) to short strides (68.3 (1.8)%; $p < 0.01$), as mean stride length decreased 19% from 137.9 (8.9) cm to 111.2(8.4) cm, respectively ($p < 0.01$).

EMG for pericapsular muscles during walking with natural self-selected and short strides are presented in Fig. 1. During natural self-selected walking, the grand ensemble average for iliocapsularis demonstrates two bursts of EMG across the gait cycle. Individual participant ensembles (Fig. 2), showed a consistent burst of EMG around toe-off in all participants, which initiated just prior to toe-off and reached peak amplitude in the early period of swing phase. Almost half (46%) of participants demonstrated an additional burst in mid-stance (Fig. 2C, D). During short strides, Iliocapsularis EMG amplitude was significantly less between 0–30% of the gait cycle ($p < 0.01$) compared to natural stride, and was significantly greater between 30–60% ($p = 0.03$) (Fig. 1). Individual participant ensembles for short strides, demonstrated 54% of participants had two bursts of iliocapsularis EMG (Fig. 2B,C), and 31% had a single burst of EMG during swing (Fig. 2A,D).

Iliacus EMG demonstrated a large burst of activity around toe-off during self-selected strides, which reached peak amplitude in early swing. During shortened strides, Iliacus EMG was less compared to natural stride in early stance (0–30%) ($p < 0.01$; Table 2), but was no different at any other point of the gait cycle. Anterior gluteus minimus EMG demonstrated a biphasic pattern of EMG during stance, with bursts in early (0–30%) and late (30–60%) stance phase. When walking with short strides, EMG amplitude was significantly greater than self-selected walking during early stance ($p = 0.03$). There was no difference between tasks during the remainder of the gait cycle. Rectus femoris EMG was characterised by a burst of EMG early in stance phase and a smaller second EMG burst at terminal stance. EMG amplitude during the initial burst was significantly less during walking with short strides than the self-selected task ($p < 0.01$), with no difference in any other part of the gait cycle (Table 2).

Anterior gluteus minimus EMG was significantly less variable between repetitions within participants, during early stance in short stride than the same period in self-selected strides (ES = 0.58, $p = 0.03$) (Supplementary Table 1). There was no significant difference in intra-participant EMG variability between the two stride length conditions for any other muscles or phases of the gait cycle.

4. Discussion

The pattern of Iliocapsularis EMG in natural walking showed a consistent burst that commenced prior to toe-off and peaked in early swing phase. There were inconsistent bursts of iliocapsularis EMG in mid to late stance. During short strides, this burst became more consistent with larger EMG amplitude in mid to late stance.

The results of our study are inconsistent with a hypothesised role of iliocapsularis to actively support the Iliofemoral ligament and control anterior hip translation in natural self-selected walking. In peak hip extension at terminal stance [12,22], the Iliofemoral ligament becomes taught. The alignment and proximity of iliocapsularis to this ligament was thought to provide active support to constrain anterior femoral head translation [1]. Taking electromechanical delay into account, iliocapsularis EMG would be expected to peak in advance of terminal stance to perform this protective role. Instead, peak EMG occurred during swing. If the primary role of iliocapsularis was active support of the Iliofemoral ligament in greater ranges of extension, its EMG would be expected to *reduce* when hip extension is reduced with shortened strides. Our observation of greater iliocapsularis EMG during mid-late stance phase of shortened strides suggests a different role. These findings do not preclude the capacity for iliocapsularis to contribute to Iliofemoral ligament support, in other populations. Observations of iliocapsularis hypertrophy in dysplasia is theorised to relate to increased muscle activity when passive stability is compromised [10]. The asymptomatic participants in the present study may already have sufficient stability provided by passive structures, limiting the necessity for iliocapsularis to actively contribute to hip joint stability. To evaluate whether iliocapsularis activity supplements joint stability in pathology, further investigation in populations with passive deficiencies, such as dysplasia would be helpful. Greater and more consistent iliocapsularis EMG in mid to late stance is more aligned to a role in tensioning the anterior joint capsule for initiation of hip flexion. Hip flexion in gait is thought to be initiated by stored energy from anterior passive structures rather than flexor muscle activity [7]. As the hip moves into extension in late stance, anterior capsular tension increases and contributes to energy storage. Commencement of activity of all muscles around toe-off and peak activity in mid swing implies a contribution to maintenance of hip flexion through swing, after passive initiation. Taken together with reduced hip extension, and thus reduced capsular tensioning during shortened strides, the greater burst of iliocapsularis EMG in mid to late stance of short stride, which was not observed for the other anterior muscles, could indicate a role providing tension to the capsule in order to augment capsular energy storage.

Activation of iliacus, anterior gluteus minimus and rectus femoris were similar to previous observations [4,5,23]. In self-selected strides; iliacus demonstrated an EMG burst in early swing, rectus femoris had

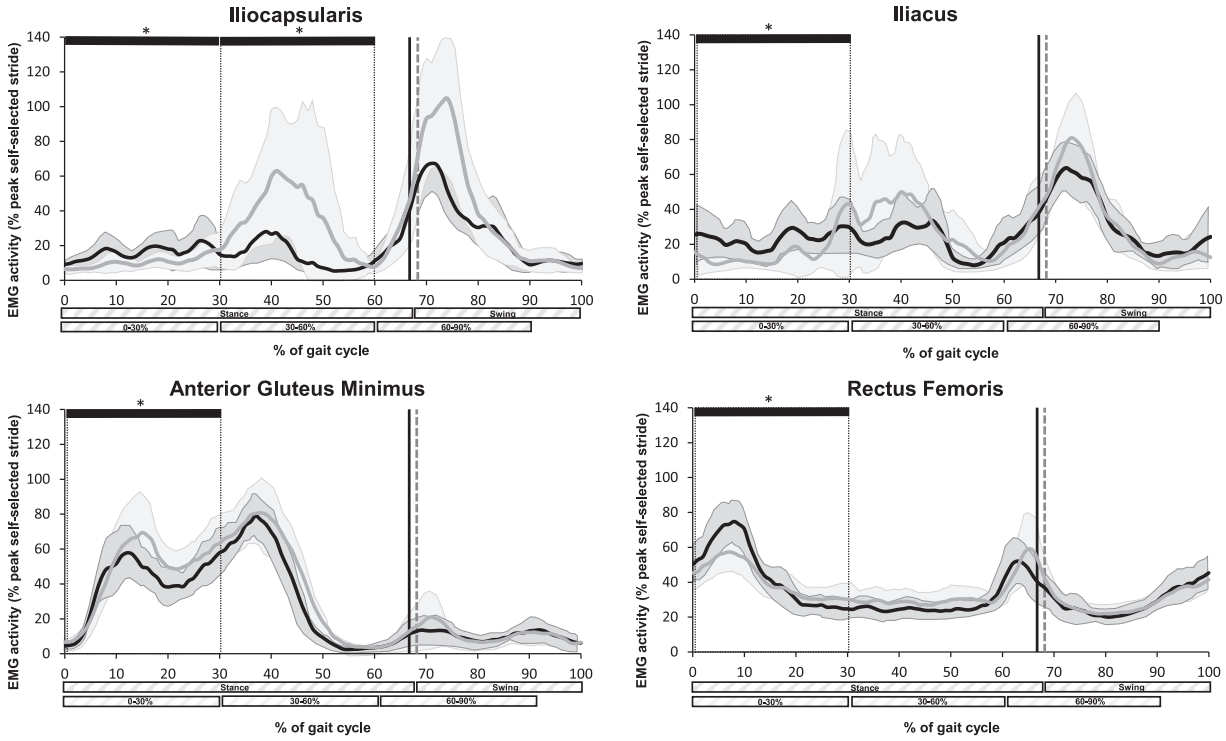

Fig. 1. Grand emsemble of electromyographic recordings, for iliocapsularis (n = 13), iliacus, anterior gluteus minimus and rectus femoris (n = 14). A comparison of self-selected (solid black line) and short stride lengths (solid grey line) throughout the gait cycle (normalised to peak EMG within each muscle during self-selected walking). Grey shaded bands indicate the 95% confidence interval (CI), light grey for short stride and darker grey for self-selected walking. Toe-off for self-selected (solid black line), and short stride (dashed grey line). Phases of the gait cycle for specific analysis highlighted by light grey horizontal bar, with diagonal lines. Statisically significant differences between short and self-selected strides are highlighted by horizontal lines at the top of the graph and dotted lines. Abbreviations; Ant GMin, anterior gluteus minimus; %, percentage; *, significant difference between self-selected and short strides.

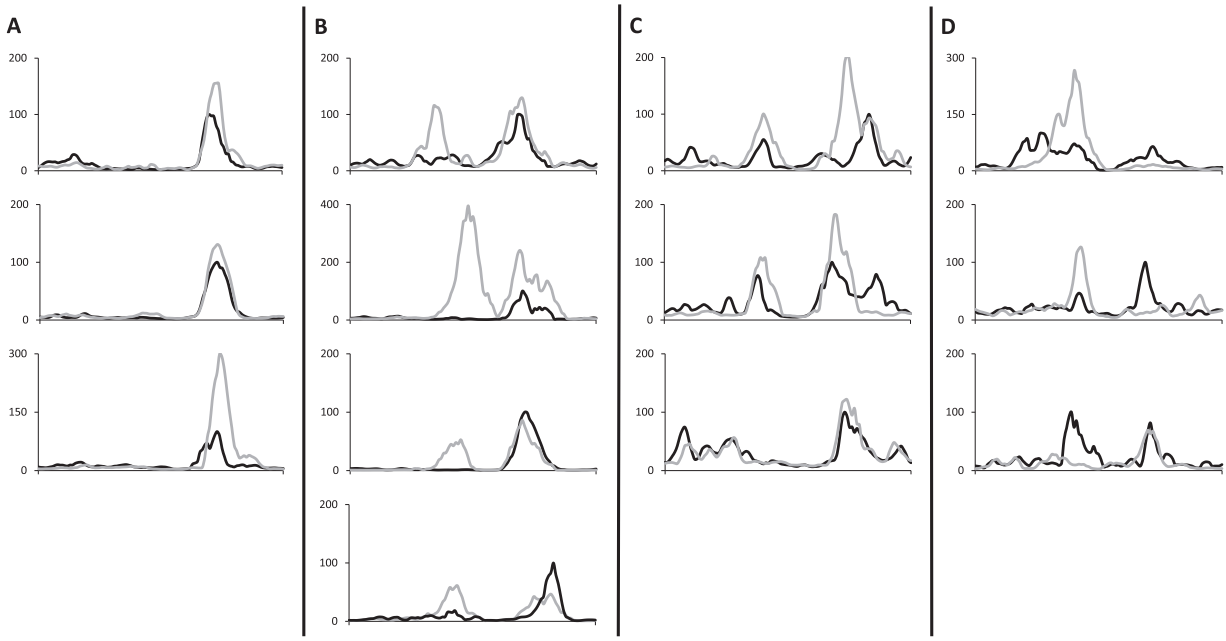

Fig. 2. Individual participant comparsion of self-selected (black line) and short stride (grey line) burst patterns for iliocapsularis (n = 13). Data for each individual are normalised to the peak activity across the gait cycle when walking at a self-selected speed (%) (vertical axis). Duration is time normalised to one hundred percent of the gait cycle (horizontal axis). Data are clustered into individuals who demonstrate; A, a single burst of EMG in swing in both self-selected and short strides; B, a single EMG burst in swing with self-selected stride and a burst in stance and swing in short strides; C, a burst in both stance and swing in self-selected and short strides; and D, a mixed burst pattern in self-selected and short strides.

Table 2
Comparisons of anterior hip muscles in self-selected and short strides across the gait cycle.

Muscle	Phase (% of gait cycle)	Normalised to peak (% of peak activity), Median (IQR)		Effect size	*p*-value*
		Self-selected Stride	Short Stride		
Iliocapsularis	0–30	10.26 (6.68–12.15)	6.18 (5.07–7.91)	0.86	< **0.01**
	30–60	10.15 (3.21–13.94)	19.02 (9.02–27.95)	−0.59	**0.03**
	60–90	26.06 (23.72–26.93)	35.44 (18.10–39.90)	−0.39	0.15
	Stance	12.16 (6.48–15.05)	13.96 (8.42–19.70)	−0.49	0.07
	Swing	21.27 (17.56–23.69)	25.48 (15.56–32.35)	−0.26	0.35
Iliacus	0–30	13.38 (7.93–17.71)	6.69 (5.12–10.10)	0.75	< **0.01**
	30–60	12.75 (9.67–17.93)	14.44 (6.31-30.32)	−0.31	0.25
	60–90	28.15 (23.50–31.78)	29.50 (19.16–35.16)	−0.36	0.18
	Stance	13.20 (10.17–18.02)	10.40 (7.74–19.95)	−0.06	0.83
	Swing	25.24 (20.19–31.00)	25.03 (12.91–30.98)	0.09	0.73
Anterior Gluteus Minimus	0–30	33.56 (21.66–39.05)	36.92 (24.16–49.04)	−0.59	**0.03**
	30–60	29.62 (26.28–34.76)	31.98 (26.06–42.21)	−0.24	0.36
	60–90	6.20 (3.47-10.45)	5.16 (3.33–8.45)	0.16	0.55
	Stance	28.07 (24.12–31.07)	31.48 (22.74–39.70)	−0.38	0.16
	Swing	7.17 (4.65–11.04)	6.98 (5.21–9.59)	−0.08	0.78
Rectus Femoris	0–30	35.92 (26.50–47.10)	32.99 (18.40–41.27)	0.81	< **0.01**
	30–60	17.81 (15.54–22.94)	18.53 (15.02–27.53)	−0.24	0.36
	60–90	25.10 (17.73–27.10)	25.41 (17.00–32.13)	−0.14	0.59
	Stance	27.02 (23.31–34.91)	28.04(19.12-35.58)	0.18	0.51
	Swing	20.18 (17.16–28.94)	23.71 (15.35–31.54)	−0.08	0.78

* Non-parametric test (Wilcoxon Sign-rank test); **Bold** = significant result. Abbreviation: IRQ interquartile range. Negative effect size indicates higher activity with short strides.

two bursts, one in early stance and another around toe-off, and anterior gluteus minimus demonstrated two EMG bursts during stance. When walking with short strides, gluteus minimus EMG was greater in early stance (0–30%), whereas iliacus and rectus femoris EMG was less, with no difference in activity observed approaching terminal stance or at any other point in the gait cycle.

Increased stride length and velocity induce greater anterior thigh muscle activity [12,24,25]. In contrast, walking with shorter strides reduced cadence, which was expected to reduce amplitude of muscle activity [26,27]. In shortened strides; walking speed is lower, the stance limb is more vertical in early stance, and there is a reduced requirement for braking force [12], with less muscle activity required to provide vertical support [24]. Our observations of decreased activity of iliacus, iliocapsularis and rectus femoris EMG during early stance (0–30%) in shortened strides supports this hypothesis. Unexpectedly anterior gluteus minimus EMG was greater during early stance (0–30%). In shorter strides, there is greater lateral centre of mass displacement [28], and less medio-lateral pelvic acceleration [26,29]. Lateral centre of mass displacement is greatest approaching mid-stance (30% of the gait cycle) [28]. Therefore, greater anterior gluteus minimus EMG may reflect an increased demand to control medio-lateral movement in this portion of the gait cycle when walking with shorter strides [26,30]. The larger burst of anterior gluteus minimus EMG in early stance is consistent with observations of muscle activity in individuals with hip pathology [14]. Habitually shortened strides are a common clinical feature in hip pathology. Taken together with the alterations in EMG identified across all anterior pericapsular muscles during early stance, walking with shortened strides may have implications for muscle action in these pathological groups. Further investigation of pericapsular action in pathological populations, may shed light on whether alterations in activity relate more to changes in stride characteristics or as a response to pain or pathology. This knowledge could help inform future clinical interventions in pathological populations, such as gait re-education.

4.1. Limitations

Several limitations require consideration. As shortened strides decrease speed [29], reduce joint range of motion [12] and reduce hip joint forces, it is impossible to isolate which variable(s) explains the differences in EMG between natural and shortened stride gait. Decreased hip joint extension was inferred from decreased stride lengths and not directly measured. Variation in how hip extension adapted with the stride length change could contribute to the variation in muscle activity observed during shortened strides, but this was not possible to test in the current study. The location of intramuscular electrodes within iliocapsularis may have differed slightly between individuals, and possible intra-muscle differences in function might explain some of the variability of activity. There was no radiological investigation of participants to exclude articular pathology. Although participants where asymptomatic it is possible that underlying articular pathology may have influenced the observed patterns of activity.

5. Conclusion

Iliocapsularis demonstrates a consistent burst of EMG around toe-off in natural walking, with inconsistent activity observed in mid-late stance. In shortened strides, the EMG burst in mid to late stance became more consistent and increased amplitude, suggesting a role for iliocapsularis in augmenting capsular tension to aid hip flexion rather than a primary role in control of anterior femoral head translation during gait. Shortened stride walking altered activity in all anterior pericapsular muscles. The relevance of these findings in clinical populations, especially those with compromised anterior hip joint control or habitually shortened strides, requires investigation.

Conflict of interest statement

The authors have no affiliations or interests that would present as a conflict of interest or allow for any bias in the preparation and reporting of this study.

CRediT authorship contribution statement

Peter Lawrenson: Conceptualization, Methodology, Validation, Formal analysis, Investigation, Data curation, Writing - original draft, Writing - review & editing, Visualization, Project administration. **Paul Hodges:** Conceptualization, Writing - review & editing, Visualization, Resources, Supervision. **Kay Crossley:** Conceptualization, Writing - review & editing, Resources, Supervision, Funding acquisition. **Bill Vicenzino:** Conceptualization, Writing - review & editing, Visualization, Supervision. **Marnie McGorm:** Conceptualization, Investigation, Writing - review & editing. **Adam Semciw:** Conceptualization, Methodology, Investigation, Resources, Validation, Formal analysis, Writing - review & editing, Visualization, Supervision, Funding acquisition.

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Appendix A. Supplementary data

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