1 # Passive elastic contribution of hip extensors to joint moments

2 # during walking in people with low back pain

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17 Manuscript word count: 4199

18 Abstract word count: 246

19

20 # Abstract

21 *Background.* It has been found that alterations in passive muscle properties may be

22 associated with low back pain, and these may be responsible for the altered gait

23 parameters often observed in subjects with back pain. The purpose of the present study

1   was to assess total hip and passive hip extensor moments in people with or without low

2   back pain during the hip flexion component of walking.

3   *Methods.* 52 subjects volunteered for this study (low back pain group, $n = 25$ (male $n =$

4   13, female $n = 12$), control group,$n = 27$ (male $n = 15$, female $n = 12$)). Passive hip

5   moments were calculated using an adapted force transducer during supine testing. A

6   biomechanical model and predictive equation were used to calculate passive hip

7   moments during walking. Total hip moments were calculated with the use of a 9 camera,

8   3-D motion-capture system.

9   *Findings.* Independent samples t-tests demonstrated no significant differences between

10   groups for gait parameters or hip or knee angles. Results of the ANOVAs demonstrated

11   significant differences in passive hip flexor moments during the second half of hip flexion

12   ($P < 0.05$).There were also significant differences in hip power and work done during

13   peaks of power absorption and the second peak of power generation ($P < 0.05$).

14   *Interpretation.* The present data demonstrates that subjects with low back pain have

15   altered passive hip extensor and total power and work done during walking compared

16   with healthy controls. Biomechanical models should include individual measurements of

17   passive joint moments.

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19

20   Passive elastic contribution of hip extensors to joint biomechanics during

21   walking in people with low back pain

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23   1   Introduction

1

2 Low back pain (LBP) is one of the leading causes of disability globally [1]. Clinical

3 assessments of LBP patients often include tests of hip extensor extensibility [2]. Tests

4 can include assessment of knee extension angle and sacral angle, the sit and reach test

5 [3], and the straight leg raise test [3, 4]. However, the relevance of hip muscle extensibility

6 to LBP and any relationship to movement remains unclear [5]. There is a growing interest

7 in including more comprehensive assessments of joint and muscle resistive properties,

8 due to the lack of consensus with assessing extensibility alone [6, 7].

9   Joint passive resistance is a property of the non-contractile tissues, such as the

10 tendon, sarcolemma, endomysium, perimysium and epimysium [6, 8-10], structural

11 proteins such as titin [11], and inactive muscle fibres. Active resistance is a property of

12 contracting muscle fibres. Some investigators report musculotendinous extensibility

13 being moderately related to passive stiffness, and weakly related to active stiffness [9].

14 An investigation by Halbertsma *et al.*[7], reported an association between hamstring

15 extensibility and LBP, but no differences in passive stiffness between LBP patients and

16 controls. Overall, there is a lack of agreement in the literature as to whether passive

17 muscle resistance is related to extensibility [9, 12]. Further, any relationship between

18 extensibility, passive and active stiffness and activities of daily living (ADLs) in LBP

19 patients remains to be determined.

20   It has been reported that muscle resistive properties may be altered in subjects

21 with LBP [6, 13, 14], and these could be responsible for the reduced leg swing, step length

22 and gait velocity often observed in LBP subjects when compared with healthy controls

23 [15-18]. During the late swing phase of walking, the activation of biceps femoris is

1   increased in LBP [18], indicating altered active factors. Identifying alterations to passive

2   or active components may be a useful tool for the clinical assessment of LBP, and the

3   first stage in developing effective physical therapy-based treatment strategies.

4   The purpose of the present study was to assess total hip and passive hip extensor

5   moments in people with LBP during the hip flexion component of walking, and to compare

6   them with pain-free controls. Passive hip moments were calculated as a product of hip

7   and knee angle using a dynamic biomechanical model. Further comparisons were made

8   of total hip power and work done during hip flexion and the complete gait cycle.

9

10   2   Methods

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12   2.1   Participants

13

14   Fifty-two subjects volunteered for this study. Subjects were excluded if they were

15   pregnant or had any tumours, rheumatological or musculoskeletal disorders, tuberculosis,

16   or an injury or infection of the spine, hips or knees during the 3 months prior to their

17   participation. Subjects were also excluded if they had a history of any dislocation or

18   surgery of the spine or lower limbs. Female subjects were only eligible for testing during

19   the 7 days following the first day of menstruation, to control for any potential effects of the

20   ovarian cycle.

21   Subjects meeting the inclusion criteria were divided into groups according to if they

22   had suffered with chronic, non-specific low-back pain (LBP group, $n = 25$ (male $n = 13$,

23   age = 34 (SD 8.53) years female $n = 12$ age= 30 (SD 7.96) years) for at least 6 weeks,

1   including at least one episode during the week of the study, or were back-pain-free (NBP

2   group, n = 27 (male n = 15, age=29 (SD 7.78) years, female n = 12, age=33 (SD 8.78)

3   years). Subjects in the NBP group needed to have been without back pain during the 6

4   months prior to the study.

5   Following consent to participate, subjects were required to complete a medical

6   screening form and International Physical Activity Questionnaire (short form) (IPAQ-SF).

7   LBP subjects were required to complete a Roland Morris Disability Questionnaire

8   (RMDQ) and to rate their level of pain on a visual analogue scale (VAS). The study was

9   approved by the ethics committees of both the University of Roehampton and the British

10   College of Osteopathic Medicine.

11

12   2.2   Experimental Setup

13

14   Passive hip extensor moments were calculated during leg raising tests using an adapted

15   force transducer, comprising a bi-axial cantilever load cell (QLA263, Futek, Irvine, US),

16   and 2 analogue electro-inclinometers (PTAM27, ASM, Moosinning, Germany). The force

17   transducer was inserted into a custom-built ankle brace designed to house the transducer

18   with minimal friction, whilst maintaining the ankle in neutral. Four knee braces were pre-

19   formed to secure the knee at 180, 170, 160 and 140 degrees, where 180 degrees refers

20   to the knee in full extension. An additional two electro-inclinometers were secured using

21   straps to the thigh and shank, to measure hip angle and angular acceleration, and knee

22   angle, respectively. Two single differential surface electromyography (EMG) electrodes

23   (SX230, Biometrics, Newport, UK), were placed over the biceps femoris and rectus

1 femoris, in accordance with the SENIAM guidelines for electrode placement. The EMG

2 signals were used for real-time feedback to ensure no activity, and not for subsequent

3 analysis.

4 The analogue signals from the load cell were pre-amplified (CSG110, Futek, Irvine,

5 US) with 15 VDC for each input and analogue-digital converted. All analogue signals

6 from the load cell and inclinometers were acquired at 50 Hz, and from the EMG electrodes

7 at 1000 Hz, using a data acquisition unit (Datalink, DLK900, Biometrics, Newport, UK).

8 Load cell and inclinometer data was digitally filtered at 2 Hz using a low-pass Butterworth

9 filter, and saved to a personal laptop computer (Dell Precision, M4500, Dell, Bracknell,

10 UK) for processing with Matlab programming software (Version 7.3, Mathworks, Natick,

11 US).

12 Subjects were required to lie supine on a massage table. In accordance with the

13 procedures of Lee and Munn [19], the test leg was passively raised 10 times to

14 precondition the tissues, and to account for variability in activity levels between subjects

15 immediately prior to testing (the test set-up is shown in figure 1). Supine passive leg

16 raises were performed 3 times with each of the 4 knee braces, with 1 minute rest between

17 tests with the same knee brace, and 2 minutes between different braces. During testing

18 the subject was required to verbally indicate if and when they felt an onset of stretch-

19 related pain. Each test would cease upon the onset of pain or deviations in EMG muscle

20 activity above resting baseline level.
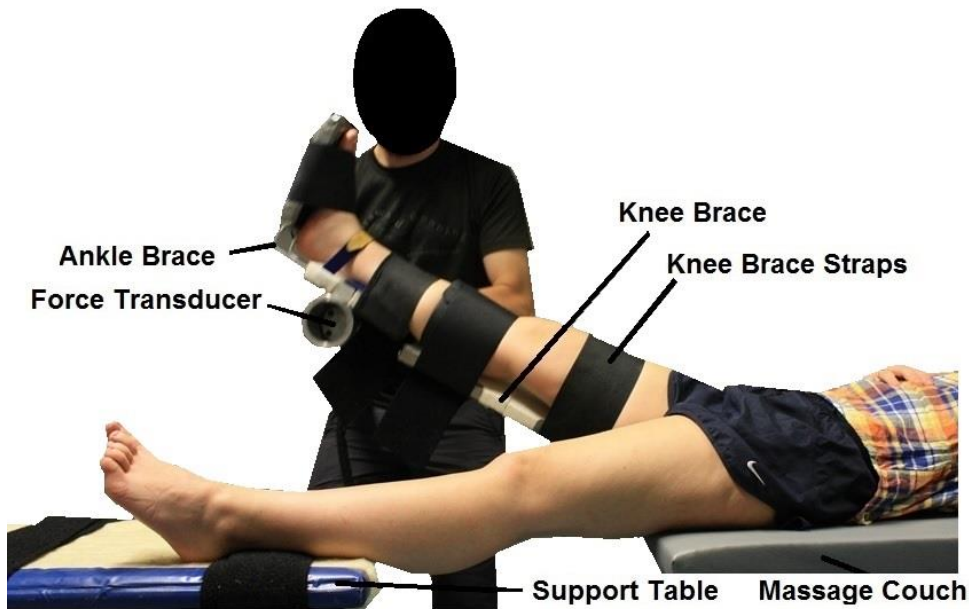
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2 Figure 1. Subject test set-up for passive hip moment assessment, showing placement of

3 force transducer, ankle brace and one of the 4 knee braces used to maintain the knee at

4 a predetermined angle (180 degrees in the example above).

5   Total hip moments were calculated during level walking with the use of a 9 camera,

6 3-D motion-capture system (T-series, Vicon, Oxford, UK) and 2 force plates (9281CA,

7 Kistler, Winterthur, Switzerland). 35 retro-reflective markers were placed on each subject

8 in accordance with the placements used by previous researchers [20-22]. Motion capture

9 data was sampled at 100-Hz and force plate data at 1000-Hz. Data was initially assessed

10 via the Nexus software programme (Vicon Nexus version 1.8, Vicon, Oxford, UK), before

11 being transferred to Microsoft Excel (2010, Microsoft, Redmond, US) and Matlab for

12 further processing.

13   Subjects were required to walk along a 10-metre walkway in view of the motion

14 capture cameras and over the force plates at their normal walking speed. The tester

15 observed foot contact onto the force plates and ensured a minimum of 5 walks contained

1   both left and right clear foot strikes. Walking was at each subject's normal pace without

2   reasonable likelihood of fatigue during testing.

3

4   2.3   Data Processing

5

6   A data processing pipeline was created in Vicon Nexus to perform standard data

7   modelling of the walking trials. The pipeline included Woltring filtering and gap filling.

8   Following determination of marker trajectories data was smoothed using a low-pass

9   Butterworth filter at 6 Hz. Total moment-angle data for the hip was established for each

10 complete gait cycle, and data was used from the first 5 complete tests on each leg. The

11 coefficient of multiple correlation (CMC) of the moment-angle curves were calculated.

12 The coefficient of variation (CV) was also calculated to assess intra-subject gait cycle

13 variability.

14 Passive hip extensor moments were calculated based upon the dynamic biomechanical

15 model developed by Lee and Munn [19]:

16

17 $$M = F_x y_f + F_y x_f + m_{leg}\ddot{x}_{cg}y_{cg} - (m_{leg}g + m_{leg}\ddot{y}_{cg})x_{cg} - m_{leg}k^2\ddot{\theta}$$

18

19 where F_x, F_y, are the forces applied to the leg to flex the hip joint, X_f, Y_f, are the locations

20 of force application to the leg, m_{leg} is the mass of the leg, g, is acceleration due to gravity,

21 x_{cg}, y_{cg}, refer to the location of the centre of mass of the leg and k is the radius of gyration.

22 $\ddot{x}_{cg}, \ddot{y}_{cg}$ refer to the acceleration of the leg centre of mass, and $\ddot{\theta}$ is the angular

1   acceleration of the leg. Dempster [23] body segment parameter ratios were used for both

2   the total and passive moment calculations.

3   Passive hip moments are a property of the length of moment arm and the

4   amount of soft tissue, and so data was normalised to body mass and height. Hip moment-

5   angle data was originally determined from 0 degrees (hip neutral) to maximum hip range

6   of motion. Because the number of samples was reduced as hip angle increased, the

7   range 0 to 60 degrees was used for analysis as a majority of subjects achieved this. The

8   mean curve from 3 leg raises, smoothed using cubic spline interpolation and fitted with

9   an exponential function, was used for further analysis.

10   To establish a predictive equation for passive hip extensor moments based upon

11   a combination of hip and knee angles, 3-D surface plots were generated for assessing

12   the changes in hip moments with hip angle, and at different knee angles. In agreement

13   with other researchers [19], passive hip moments were found to increase exponentially

14   as a function of hip angle. By introducing a variable knee angle component, it was found

15   that this contributed a linear component to the equation. A surface fitting programme was

16   written for Matlab, and visual and residual analyses used to determine the most

17   appropriate equation coefficients for the individual datasets.   From the present

18   investigation, the following equation was developed to predict passive hip moments

19   ($M_{passive}$) during flexion:

20

21                $$M_{passive} = a * \theta_{knee} + b * exp^{c*\theta_{hip}} + d$$

22

1   where θ_{knee} is the angle at the knee, θ_{hip} the angle at the hip, and *a*, *b*, *c*, and *d* are the

2   equation coefficients.

3   Passive hip moments and 95% confidence intervals were recalculated from hip

4   neutral to maximum hip flexion angle, using the derived predictive equation. Root mean

5   squared error (RMSE) and adjusted r-squared were calculated to assess goodness of fit

6   of each surface-plot.

7   Hip moments were calculated using the conventional gait model, based upon the

8   Newington-Helen Hayes model. Total hip power was calculated from hip angular velocity

9   and total hip moments.   All data was subsequently normalised to body mass and height.

10 Total hip moments were calculated at the peak hip flexor moment (FL), and the two hip

11 extensor peaks (Ext1, Ext2). Total hip power was calculated at the two peaks of power

12 generation (H1, H3) and peak of power absorption (H2). Mechanical work done at H1

13 and H3 was calculated by integrating the complete positive portions of the corresponding

14 power generation curves. Work done was also calculated for the full negative power

15 absorption curve (H2). Passive hip extensor moments and total hip moments, power and

16 work done were additionally calculated for the portion of the gait cycle corresponding to

17 hip flexion, between hip neutral and maximum hip flexion angle. Scores for the RMDQ

18 were calculated as the sum of ticked statements. Physical activity data was used to

19 ensure matching of subjects in LBP and NBP groups.

20

21 2.4   Statistical Analysis

22

1   A mixed-model analysis of variance (ANOVA) was used to compare within gait cycle

2   peaks and degrees of hip flexion between LBP and NBP group data using SPSS (version

3   24, IBM Statistics, U.S.). *Post hoc* analyses for between groups comparisons were

4   performed using independent samples t-tests where significant interactions were

5   determined. Passive hip extensor moments and total hip moments, power and work done

6   were compared at 25% increments of hip flexion, between neutral and maximum hip

7   flexion. Total hip moments were additionally compared at FL, Ext1 and Ext2, and total

8   power and work done were compared at H1, H2 and H3. Additional comparisons included

9   age, gender, body mass, physical activity type and duration, walking speed and step

10 length, maximum hip flexion angle, hip extension angle and knee angle, assessed using

11 independent t-tests.   CMCs and CVs were used to assess intra-subject gait cycle

12 characteristics. RMSE and the adjusted r-squared were used to assess goodness of fit

13 of the predictive equation coefficients for calculating passive moment contributions.

14 Normality of data was assessed using Shapiro-Wilk tests, and the alpha level for all tests

15 was set at 0.05.

16

17 3   Results

18

19 Subjects in the LBP group reported occurrence of back pain for 6.97 (SD 5.98) years with

20 a range of 6 weeks to 30 years. RMDQ scores were 3.92 (SD 3.1) and VAS scores were

21 5.06 (SD 2.27). Independent samples t-tests demonstrated no significant differences

22 between LBP and NBP groups for age, gender, body mass (within gender), walking speed

23 or step length ($P > 0.05$). There were no significant differences ($P > 0.05$) in any of the

1 physical activity parameters calculated, which included walking (13.6 (SD 18.3) hours per

2 week LBP, 9.9 (SD 12.9) hours NBP), moderate intensity exercise (6.9 (SD 11.0) hours

3 per week LBP, 4.1 (SD 5.6) hours NBP) and vigorous exercise (6.7 (SD 8.7) hours per

4 week LBP, 5.1 (SD 3.4) hours NBP).

5   The CMC means were 0.955 (SD 0.037) for LBP and 0.937 (SD 0.051) for NBP,

6 respectively. The CV means were 21.4% (SD 11.6) and 26.3% (SD 13.9), for LBP and

7 NBP, respectively. The RMSE was 0.016 (SD 0.008) Nm/(kg*m) for LBP and 0.017 (SD

8 0.07) Nm/(kg*m) for NBP. The adjusted r-squared values were 0.889 (SD 0.081) for LBP

9 and 0.896 (SD 0.075) for NBP. There were no significant differences in hip or knee angles

10 between groups ($P > 0.05$).

11   There were no statistically significant interaction effects in the total hip flexor (FL)

12 or extensor moment peaks (Ext1, Ext2) ($F_{(1.571,155)} = 3.019, P = 0.065$, table 1) or their

13 timings ($F_{(1.599,158)} = 0.368$, $P = 0.645$) (figure 2). There was no main effect of group

14 $F_{(1,99)}=0.208$, $P = 0.649$).   During the hip flexion component of the gait cycle,

15 there was no statistically significant interaction effect in total hip moments ($F_{(2.064,204)}$

16 $= 0.823$, $P = 0.444$) (table 2). There was no main effect of group ($F_{(1,99)} = 0.012$, $P =$

17 0.913).

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Hip Moments with 95% Confidence Intervals

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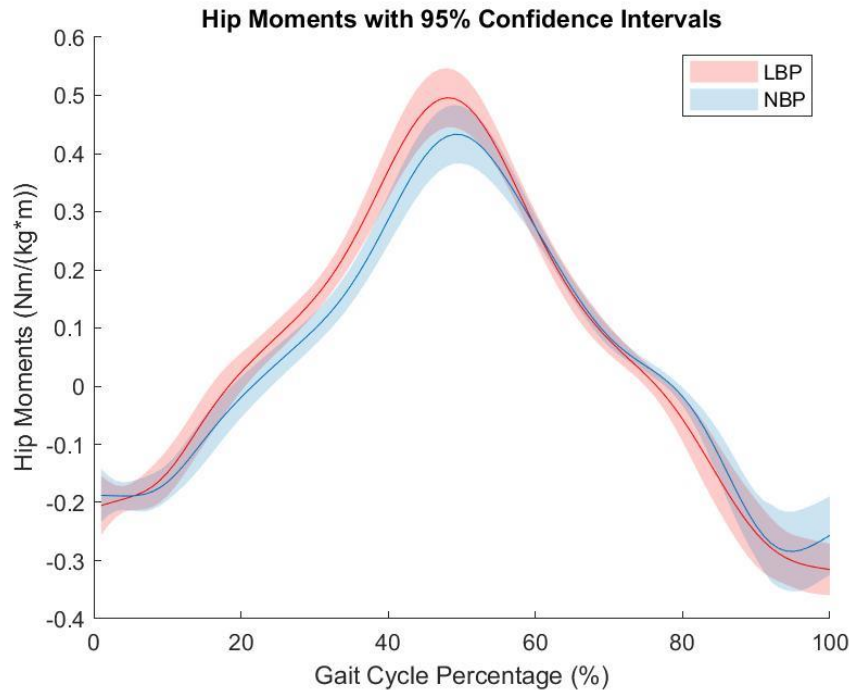
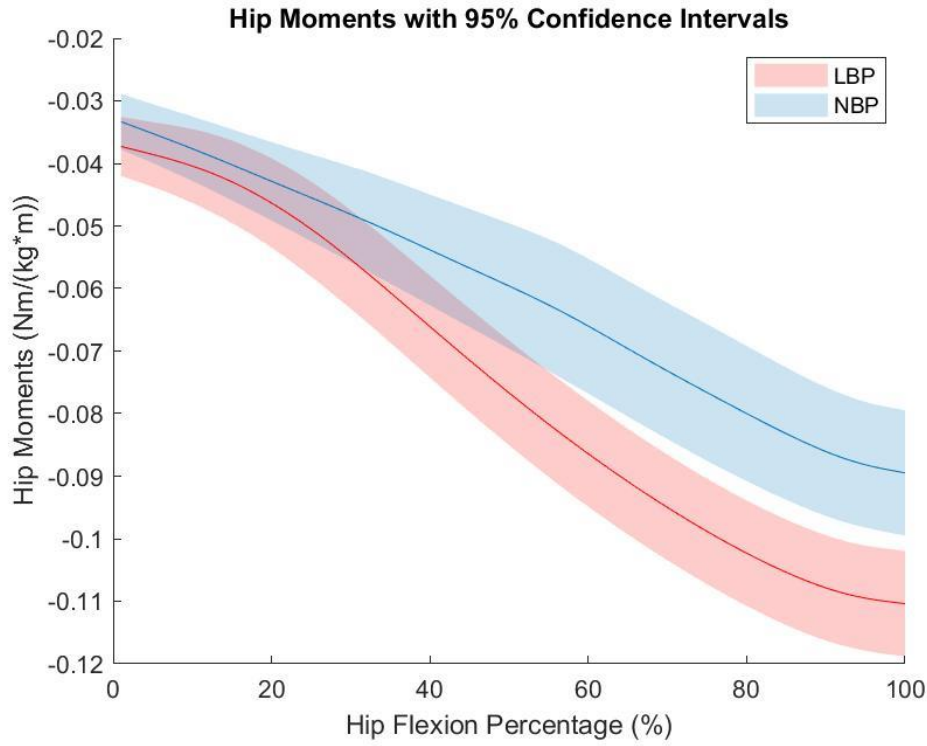Figure 2. Mean total hip moments with 95% confidence intervals in LBP and NBP groups during gait cycle.

For passive hip moments there was a significant interaction effect of group and percentage of hip flexion ($F(1.751,173)$ = 9.316, P = 0.000) and a statistically significant main effect of group ($F(1,99)$ = 6.597, P = 0.012). Passive hip moments decreased throughout hip flexion, demonstrating increased extensor moments (figure 3), with no significant differences between groups at hip neutral or 25% ($P > 0.05$). From 50-100% of hip flexion, passive moments were significantly more negative in LBP subjects ($P < 0.05$), demonstrating greater extensor moments. At maximum hip flexion, passive hip extensor moments were 46.6% and 39.4% of total hip moments, for LBP and NBP, respectively. Passive hip extensor and total hip moment mean, standard deviation and results of the independent t-tests are shown in table 2.

13

1 Table 1. Variable data for complete gait cycle. NS (not significant) indicates ANOVA *P* >
2 0.05 with *post hoc* tests not indicated. Statistically significant results are denoted with an
3 asterisk(*).
4

Moments (Nm/(kg*m))		Mean (SD)	Confidence Interval		ANOVA *P*-value	Pairwise *P*-value
			Lower	Upper		
FL	LBP	0.557 (0.169)	0.51	0.604		NS
	NBP	0.482 (0.174)	0.435	0.529		
Ext1	LBP	-0.317 (0.122)	-0.351	-0.282	0.065	NS
	NBP	-0.278 (0.119)	-0.31	-0.245		
Ext2	LBP	-0.434 (0.153)	-0.477	-0.391		NS
	NBP	-0.421 (0.221)	-0.481	-0.361		
Power (W/(kg*m))						
H1	LBP	0.280 (1.96)	0.225	0.335		0.99
	NBP	0.281 (0.159)	0.237	0.324		
H2	LBP	-0.520 (0.231)	-0.585	-0.456	0.027*	0.041*
	NBP	-0.429 (0.212)	-0.487	-0.372		
H3	LBP	0.844 (0.298)	0.761	0.927		0.045*
	NBP	0.736 (0.235)	0.672	0.8		
Work Done (J/(kg*m))						
H1	LBP	0.028 (0.028)	0.02	0.036		0.329
	NBP	0.034 (0.027)	0.026	0.041		
H2	LBP	-0.111 (0.059)	-0.128	-0.094	0.010*	0.034*
	NBP	-0.087 (0.054)	-0.101	-0.072		
H3	LBP	0.115 (0.033)	0.106	0.124		0.017*
	NBP	0.101 (0.026)	0.094	0.108		

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6

Hip Moments with 95% Confidence Intervals

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Figure 3. Mean passive hip extensor moments with 95% confidence intervals in LBP and
NBP groups during hip flexion from neutral to maximum hip flexion

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Table 2. Passive hip extensor and total hip moments (Nm/(kg*m)) during hip flexion, from neutral to full hip flexion. NS (not significant) indicates ANOVA *P* > 0.05 with *post hoc* tests not indicated. *P*-values included represent results of *post hoc* independent t-tests. Statistically significant results are denoted with an asterisk(*).

Hip Flexion Angle (%)			Moments (Nm/(kg*m))	Confidence Interval		Pairwise
			Mean (S.D.)	Lower	Upper	*P*-value
0	Passive	LBP	-0.037 (0.017)	-0.042	-0.033	0.225
		NBP	-0.033 (0.016)	-0.038	-0.029	
	Total	LBP	0.228 (0.083)	0.204	0.251	NS
		NBP	0.194 (0.080)	0.172	0.216	
25	Passive	LBP	-0.050 (0.027)	-0.058	-0.043	0.345
		NBP	-0.046 (0.026)	-0.052	-0.039	
	Total	LBP	0.075 (0.069)	0.056	0.094	NS
		NBP	0.074 (0.073)	0.054	0.094	
50	Passive	LBP	-0.077 (0.030)	-0.085	-0.068	0.012*
		NBP	-0.060 (0.037)	-0.07	-0.05	
	Total	LBP	0.011 (0.043)	-0.002	0.023	NS
		NBP	0.020 (0.044)	0.008	0.032	
75	Passive	LBP	-0.099 (0.03)	-0.107	-0.09	0.002*
		NBP	-0.077 (0.040)	-0.088	-0.066	
	Total	LBP	-0.141 (0.168)	-0.188	-0.094	NS
		NBP	-0.116 (0.167)	-0.162	-0.071	
100	Passive	LBP	-0.110 (0.030)	-0.118	-0.102	0.002*
		NBP	-0.089 (0.037)	-0.099	-0.079	
	Total	LBP	-0.236 (0.192)	-0.29	-0.183	NS
		NBP	-0.226 (0.257)	-0.296	-0.157	

There was a statistically significant interaction of group and peak of hip power ($F_{(1.353,134)}$ = 4.366, *P* = 0.027) but no main effect of group ($F_{(1,99)}$ = 0.06, *P* = 0.807). There were no significant interactions of angle at peak ($F_{(1.526,151)}$ = 0.863, *P* = 0.398). The first peak of power generation (H1) occurred at 8.2 (SD 4.5) %, and 10.0 (SD 4.4) % of gait cycle, for LBP and NBP, respectively, with no significant differences in timing or

1   power ($P > 0.05$. table 1, figure 4). The hips then absorbed power, with a peak of power

2   absorption (H2) during mid-stance. The peak of power absorption at H2 was significantly

3   greater in LBP than NBP (-0.520 (SD 0.231) W/(kg*m), -0.429 (SD 0.212) W/(kg*m), for

4   LBP and NBP, respectively, $P = 0.041$), with no difference in timings between groups (P

5   > 0.05).   Power generation peaked at toe-off and the initiation of swing phase, at

6   approximately 60% GC (H3). H3 was significantly greater in LBP than NBP (0.844 (SD

7   0.298) W/(kg*m), 0.736 (SD 0.235) W/(kg*m) for LBP and NBP, respectively, $P = 0.045$),

8   with no difference in timings between groups ($F(1.526,151)$, $= 0.863$, $P = 0.398$).
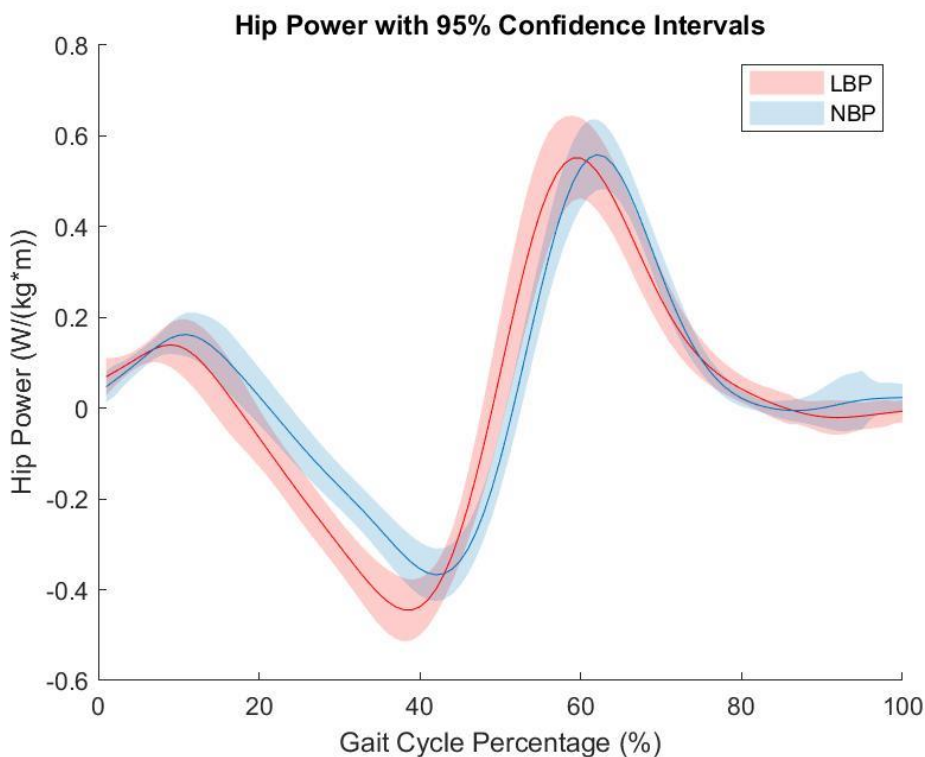

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10   Figure 4. Mean total hip power with 95% confidence intervals in LBP and NBP groups
11   during gait cycle.
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13   For power during the hip flexion component of gait (figure 5), ANOVA results

14   demonstrated a significant interaction of group and time ($F(2.264,224) = 3.228$, $P = 0.036$)

17

1   but no main effect of group ($F(1,99) = 2.538$, $P = 0.114$). *Post hoc* analyses determined

2   that total hip power was significantly greater ($P = 0.012$) in LBP subjects when the hip

3   was in neutral (LBP = 0.717 (0.300) W/(kg*m), NBP = 0.583 (0.22) W/(kg*m). There were

4   no statistically significant differences between groups at any other percentage of hip

5   flexion assessed ($P > 0.05$). Mean, standard deviation and results of the independent t-

6   tests for hip flexion are shown in table 3.

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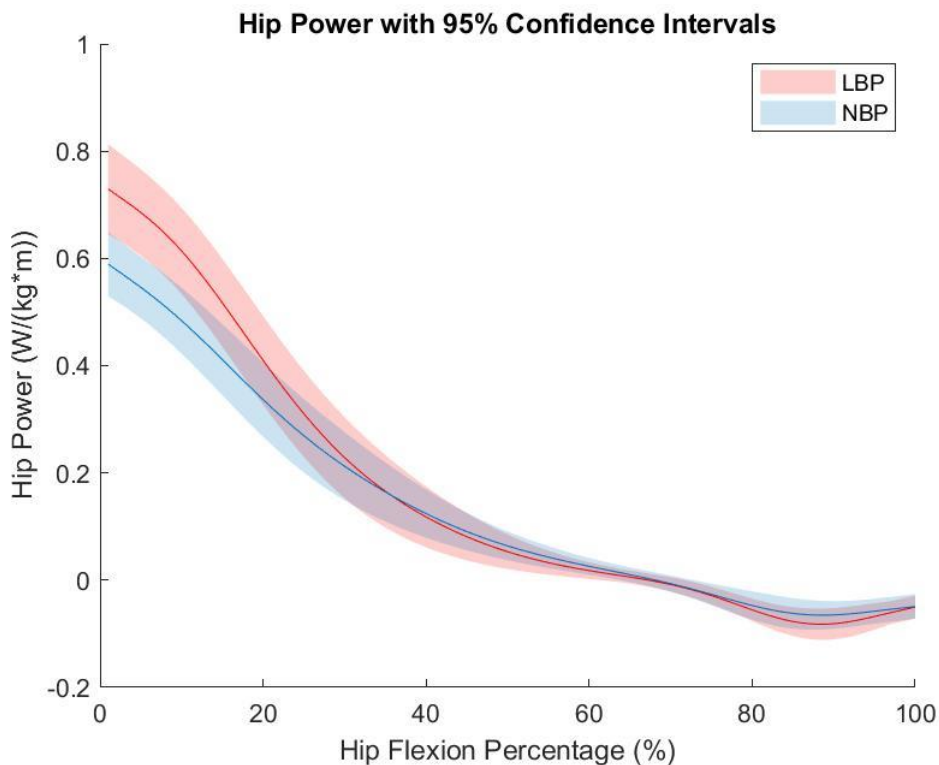
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9   Figure 5. Mean hip power with 95% confidence intervals in LBP and NBP groups during
10   hip flexion from neutral to maximum hip angle.
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14   Table 3. Total hip power (W/(kg*m)) during hip flexion, from neutral to full hip flexion.
15   Statistically significant results are denoted with an asterisk(*).

Hip Flexion Angle (%)			Power (W/(kg*m))				
			Mean (S.D.)	Confidence Interval		ANOVA	Pairwise
				Lower	Upper	*P-value*	*P*-value
0	LBP		0.717 (0.300)	0.633	0.801		0.012*
	NBP		0.583 (0.220)	0.523	0.643		
25	LBP		0.309 (0.287)	0.228	0.389		0.455
	NBP		0.269 (0.252)	0.2	0.337		
50	LBP		0.053 (0.115)	0.021	0.085	0.036*	0.608
	NBP		0.064 (0.098)	0.037	0.09		
75	LBP		-0.027 (0.059)	-0.043	-0.01		0.928
	NBP		-0.026 (0.074)	-0.046	-0.006		
100	LBP		-0.054 (0.076)	-0.075	-0.032		0.868
	NBP		-0.051 (0.085)	-0.074	-0.028		

There was a significant interaction of group and peak for work done during gait ($F(1.263,125)$ = 6.114, P = 0.010), but no main effect of group ($F(1,99)$ = 2.538, P = 0.114). Negative mechanical work done during the H2 power absorption curve was greater in LBP than NBP (-0.111 (SD 0.059) J/(kg*m), -0.087 (SD 0.054) J/(kg*m), for LBP and NBP, respectively, P = 0.034). Positive mechanical work was greater in LBP than NBP during the H3 power generation curve (0.115 (SD 0.033) J/(kg*m), 0.101 (SD 0.026) J/(kg*m), for LBP and NBP, respectively, P = 0.017). During H1 and the hip flexion component of the gait cycle there were no statistically significant differences in work done between groups (P > 0.05). Total work done mean, standard deviation and results of the independent t-tests for hip flexion are shown in table 4.

Table 4. Total hip mechanical work done (J/(kg*m)) during hip flexion, from neutral to full hip flexion in 25% intervals and for overall hip flexion

19

Hip Flexion Angle (%)			Work Done (J/(kg*m))			
			Mean (S.D.)	Confidence Interval		P-value
				Lower	Upper	
25	LBP		0.038 (0.017)	0.033	0.043	
	NBP		0.031 (0.015)	0.027	0.035	
50	LBP		0.009 (0.012)	0.006	0.013	
	NBP		0.010 (0.011)	0.007	0.013	
75	LBP		0.001 (0.003)	0	0.002	0.096
	NBP		0.001 (0.004)	0	0.002	
100	LBP		-0.006 (0.007)	-0.007	-0.004	
	NBP		-0.005 (0.007)	-0.006	-0.003	
Total Flexion	LBP		0.043 (0.028)	0.036	0.051	
	NBP		0.038 (0.028)	0.03	0.046	

4 Discussion

The present study found no statistically significant differences in hip and knee angles between groups. Values of passive hip extensor moments, and total hip moments, power and work done for NBP subjects were similar to those reported elsewhere [24]. Other investigators have reported alterations in hip and knee angles during walking in LBP subjects compared with healthy controls [25-27]. The most likely cause of increased passive moments in LBP is an increased passive resistance of the hip extensor and knee flexor muscles, such as the bi-articular hamstrings muscles. However, a previous study using the same subjects demonstrated no significant differences in passive moments between groups. Because passive hip moments increase with hip angle in an exponential

1   manner, it is likely that small increases in hip angle have a considerable influence on

2   passive moments during walking, as observed in the present study. It is also plausible

3   that individual alterations and interactions of joint angles and passive moments may be

4   masked by averaging individual and group leg data. Alternatively, the difference in

5   passive moments equates to a maximum of 3 Nm in absolute rather than normalised

6   figures, which may be statistically significant but of little, if any, clinical relevance. Altered

7   passive moments occurred without differences in total moments.

8

9   Power absorption and work done at H2 were significantly greater in LBP than NBP.

10   The H2 power curve occurs during the stance phase of gait, as the hip flexors

11   eccentrically contract to stabilise the pelvis and support the trunk, preventing backwards

12   movement of the trunk relative to the pelvis. Power generation and work done at H3 were

13   also greater in LBP than NBP, and occurs during the late stance (pre-swing) phase of

14   gait, as the hip flexors contract concentrically to pull the leg forward and facilitate toe-off.

15   This coincided with greater power generation and work in LBP than NBP as the hip moved

16   through neutral.

17   The results of the present study demonstrate increased passive extensor moments

18   in LBP, with no differences in total moments. Whether the passive moments contribute

19   to total moments, thereby reducing the need for active contraction and promoting

20   efficiency, or if values of total moments should be altered due to the influence of greater

21   passive moments, is not currently understood [24]. Alternatively, altered joint angles, and

22   therefore passive moments, later in the swing phase may follow the increased total power

23   generation from the hip flexors during early swing, directly following toe-off. In this case,

1   increased passive moments may be due to increased joint angles, resulting from greater

2   momentum of the hip where active power generation is greater in LBP with no differences

3   in passive moments following toe-off. Greater momentum of the swinging limb could

4   cause greater hip flexion angles, contributing to greater passive resistance, and therefore

5   higher passive moments. The lack of difference in total moments may reflect reduced

6   active components late in swing, or be due to total moments being miscalculated due to

7   lack of account of passive influences in inverse dynamic modelling. Despite our lack of

8   understanding regarding the mechanisms and interactions of the passive and active

9   components, it is clear the passive component is considerable, and should not be

10   overlooked in future models of dynamic movements.

11   LBP subjects in the present study reported RDMQ scores that were low (3.92 (SD

12   3.1)) compared with other studies, VAS scores were considered moderate (5.06 (SD

13   2.27)), and similar to those reported elsewhere [28-35]. Although self-reported pain is a

14   subjective measure, subjects in the studies where VAS and RMDQ scores were higher

15   were recruited from physical therapy and rehabilitation clinics [28, 29, 31]. Overall, the

16   findings in the present study remain appropriate to the population assessed. Whilst LBP

17   severity and disability may be lower than in subjects attending rehabilitation clinics, there

18   were clear adaptations in gait to modify passive and active joint biomechanics compared

19   with healthy controls.

20   The present study indicates that passive moments may be influenced without

21   affecting total joint moments. Specifically, it is plausible that LBP may affect the structural

22   components of muscle tissue, or alternatively results in subtle interactions of joint angle

23   and moments during movement, sufficient to influence passive components without

1   affecting total moment.   There may be further interactions between LBP, passive

2   moments, total power and total work done, which affect accelerations and energy

3   efficiency for movement.   From a clinical perspective, an assessment of passive

4   biomechanics will be useful to help understand which tissue types are influenced by LBP,

5   on a case-by-case basis. Findings suggestive of structural adaptations will necessitate a

6   muscle-targeted therapy, such as stretching or strengthening exercise.   Conversely, if

7   joint kinematics are altered, this indicates the active component of the muscular system

8   may have been affected, necessitating a therapy that targets neuromuscular

9   improvements, such as muscle activation timing and magnitude, and the specific fibres

10   recruited during movement.

11   A limitation of the current study was the measurement of hip passive moments

12   during hip flexion only. In previous studies [24, 36, 37], researchers averaged hip

13   moments during flexion and extension, and reported only a small underestimation of

14   moments during flexion, and overestimation during extension. However, in the present

15   study it was considered more appropriate to determine accurate values of passive hip

16   extensor moments, for integration into the hip flexion model during walking, rather than

17   over- or underestimating moments.   Future studies would benefit from direct

18   measurement of both passive hip extensor and hip flexor moments to model the complete

19   gait cycle accurately in the sagittal plane.

20   Hip extensor moments were 46.6% and 39.4% of total hip moments for LBP and

21   NBP, respectively, at maximum hip flexion angle. This finding demonstrates that passive

22   structures may interact with active components, have a considerable influence on total

23   hip moments during walking [24], and should be integrated into future biomechanical

1 models. Whether it is appropriate to add passive moments to total moments, or to

2 subtract them, or if there is a more complex interaction between passive and active

3 components to total moments is not currently understood [24] and therefore warrants

4 further investigation.

5

6 5 Conclusions

7

8 The present data demonstrates that subjects with LBP have altered passive hip extensor

9 moments and total power and work done during walking, compared with healthy controls.

10 Although it is not possible to extrapolate cause and effect relationships, rehabilitation

11 programmes for LBP patients should differentiate between the active, neuromuscular

12 components of movement, and the passive components. Biomechanical models should

13 include individual measurements of passive joint moments. The approach used in the

14 present study may be a useful measurement model for clinicians assessing low back pain.

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