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Passive elastic contribution of hip extensors to joint moments during walking in people with low back pain

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1 Passive elastic contribution of hip extensors to joint moments

2 during walking in people with low back pain

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20 Abstract

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

22 affected byassociated with low back pain (LBP), and these may be responsible for the

23 altered gait parameters often observed in subjects with back painLBP. The purpose of

1

the present study was to assess total hip and passive hip extensor moments in people with ~~or without~~ ~~LBP~~ low back pain during the hip flexion component of walking.

Methods. 52 subjects volunteered for this study (~~LBP~~ low back pain group, *n* = 25 (male *n* = 13, female *n* = 12), control group, ~~(NBP),~~ *n* = 27 (male *n* = 15, female *n* = 12)). Passive hip moments were ~~measured~~ calculated using an adapted force transducer during supine testing. A biomechanical model and predictive equation were used to calculate passive hip moments during walking. Total hip moments were ~~measured~~ calculated with the use of a 9 camera, 3-D motion-capture system.

Findings. Independent samples t-tests demonstrated no significant differences between groups for gait parameters or hip or knee angles. Results of the ANOVAs demonstrated significant differences in passive hip flexor moments during the second half of hip flexion ($P < 0.05$). ~~There were significant differences in the peak of hip flexor moments, in moments at hip neutral, and in passive moments during the second half of hip flexion (*P* < 0.05).~~ There were also significant differences in hip power and work done during peaks of power absorption and the second peak of power generation (*P* < 0.05).

Interpretation. The present data demonstrates that subjects with ~~LBP~~ low back pain have altered passive hip extensor and total ~~hip moments,~~ power and work done during walking compared with healthy controls. Biomechanical models should include individual measurements of passive joint moments.

Passive elastic contribution of hip extensors to joint biomechanics during walking in people with low back pain

1

1   Introduction

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

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

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

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

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

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

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

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

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

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

contracting muscle fibres.   Some investigators report musculotendinous extensibility

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

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

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

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

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

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

patients remains to be determined.

        It has been ~~found~~ reported that muscle resistive properties ~~can be affected by LBP~~

may be altered in subjects with LBP [6, 13, 14], and these ~~may~~ could be responsible for

1   the reduced leg swing, step length and gait velocity often observed in LBP subjects when

2   compared with healthy controls [15-18]. During the late swing phase of walking, the

3   activation of biceps femoris is increased in LBP [18], indicating altered active factors.

4   Identifying alterations to passive or active components may be a useful tool for the clinical

5   assessment of LBP, and the first stage in developing effective physical therapy-based

6   treatment strategies.

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

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

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

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

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

12

13   2   Methods

14

15   2.1   Participants

16

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

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

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

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

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

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

23   ovarian cycle.

4

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

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

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

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

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

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

7 months prior to the study.

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

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

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

11 (RMDQ) and to rate their level of pain on a visual analogue scale (VAS). ~~During the~~

12 ~~recruitment process the groups were matched for age, body mass, gender and physical~~

13 ~~activity habits (type and experience).~~ The study was approved by the ethics committees

14 of both the University of Roehampton and the British College of Osteopathic Medicine.

15

16 2.2   Experimental Setup

17

18 Passive hip extensor moments were ~~measured~~ calculated during leg raising tests using

19 an adapted force transducer, comprising a bi-axial cantilever load cell (QLA263, Futek,

20 Irvine, US), and 2 analogue electro-inclinometers (PTAM27, ASM, Moosinning,

21 Germany). The force transducer was inserted into a custom-built ankle brace designed

22 to house the transducer with minimal friction, whilst maintaining the ankle in neutral. Four

23 knee braces were pre-formed to secure the knee at 180, 170, 160 and 140 degrees,

5

1 where 180 degrees refers to the knee in full extension. An additional two electro-

2 inclinometers were secured using straps to the thigh and shank, to measure hip angle

3 and angular acceleration, and knee angle, respectively. Two single differential surface

4 electromyography (EMG) electrodes (SX230, Biometrics, Newport, UK), were placed

5 over the biceps femoris and rectus femoris, in accordance with the SENIAM guidelines

6 for electrode placement. The EMG signals were used for real-time feedback to ensure

7 no activity, and not for subsequent analysis.

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

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

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

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

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

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

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

15 US).

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

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

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

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

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

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

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

6

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

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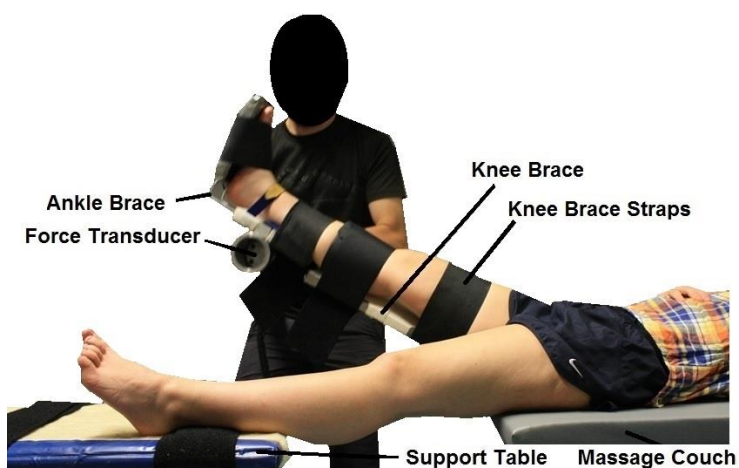
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4   Figure 1. Subject test set-up for passive hip moment assessment, showing placement of

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

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

7

8   Total hip moments were measuredcalculated during level walking with the use of

9   a 9 camera, 3-D motion-capture system (T-series, Vicon, Oxford, UK) and 2 force plates

10 (9281CA, Kistler, Winterthur, Switzerland). 35 retro-reflective markers were placed on

11 each subject in accordance with the placements used by previous researchers [20-22].

12 Motion capture data was sampled at 100-Hz and force plate data at 1000-Hz. The data

13 was stored on a personal computer (Dell Precision, M4500, Dell, Bracknell, UK). Data

14 was initially assessed via the Nexus software programme (Vicon Nexus version 1.8,

1 Vicon, Oxford, UK), before being transferred to Microsoft Excel (2010, Microsoft ~~Excel~~,

2 Redmond, US) and Matlab for further processing.

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

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

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

6 both left and right clear foot strikes. ~~Once this was achieved the tester completed~~

7 ~~preliminary processing to ensure all markers were still visible in a minimum of 5 of the~~

8 ~~walks for each leg. If this was not the case further tests were carried out until sufficient~~

9 ~~complete data was available.~~Walking was at each subject's normal pace without

10 reasonable likelihood of fatigue during testing.

11

12 2.3   Data Processing

13

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

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

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

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

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

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

20 ~~Where CMC values were less than 0.8 the moment-angle curves were visually assessed~~

21 ~~for any outlying curve, with outliers replaced by other gait cycle data for that subject, or~~

22 ~~removed and the mean and CMC recalculated based upon the remaining data.~~ The

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

2   variability.

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

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

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

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

7   ~~individual limitations in range of motion, the range 0 to 60 degrees was used for analysis~~

8   ~~as a majority of subjects achieved this.~~ Passive hip extensor moments were calculated

9   based upon the dynamic biomechanical model developed by Lee and Munn [19]:

10

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

12

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

14   locations of force application to the leg, m_{leg} is the mass of the leg, g, is acceleration due

15   to gravity, x_{cg}, y_{cg}, refer to the location of the centre of mass of the leg and k is the radius

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

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

18   the total and passive moment calculations.

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

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

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

22   of motion. Because the number of samples was reduced as hip angle increased, ~~due to~~

23   ~~individual limitations in range of motion, the range 0 to 60 degrees was used for analysis~~

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9

1   as a majority of subjects achieved this. The ~~ensemble~~ mean curve ~~from~~of 3 ~~lifts~~leg raises,

2   smoothed using cubic spline interpolation and fitted with an exponential function, was

3   used for further analysis.

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

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

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

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

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

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

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

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

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

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

14

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

16

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

18   equation coefficients.

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

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

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

22   of each surface-plot.

10

1   Hip moments were calculated ~~within the Vicon Nexus Plug-in Gait software,~~ using

2   the conventional gait model,. based upon the Newington-Helen Hayes model. Total hip

3   power was calculated ~~based upon~~from hip angular velocity and total hip moments.   All

4   data was subsequently normalised to body mass and height. Total hip moments were

5   calculated at the peak hip flexor moment (FL), and the two hip extensor peaks (Ext1,

6   Ext2). Total hip power was calculated at the two peaks of power generation (H1, H3) and

7   peak of power absorption (H2). Mechanical work done at H1 and H3 was calculated by

8   integrating the complete positive portions of the corresponding power generation curves.

9   Work done was also calculated for the full negative power absorption curve (H2). Passive

10   hip extensor moments and total hip moments, power and work done were additionally

11   calculated for the portion of the gait cycle corresponding to hip flexion, between hip neutral

12   and maximum hip flexion angle. Scores for the RMDQ were calculated as the sum of

13   ticked statements. ~~Scores for both RMDQ and VAS were entered into a Microsoft Excel~~

14   ~~spreadsheet for further analysis.~~ Physical activity data was used to ensure matching of

15   subjects in LBP and NBP groups.

16

17   2.4   Statistical Analysis

18

19   ~~Independent t-test~~ A mixed-model analysis of variance (ANOVA) was used to compare

20   within gait cycle peaks and degrees of hip flexion between LBP and NBP group data using

21   SPSS (version 24, IBM Statistics, U.S.). *Post hoc* analyses for between groups

22   comparisons were performed using independent samples t-tests where significant

23   interactions were determined. Passive hip extensor moments and total hip moments,

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11

power and work done were compared at 25% increments of hip flexion, between neutral

and maximum hip flexion. Total hip moments were additionally compared at FL, Ext1 and

Ext2, and total power and work done were compared at H1, H2 and H3.   Additional

comparisons included age, gender, body mass, physical activity type and duration,

walking speed and step lengthgait parameter data, maximum hip flexion angle, hip

extension angle and knee angle, assessed using independent t-tests. CMCs and CVs

were used to assess intra-subject gait cycle characteristics. RMSE and the adjusted r-

squared were used to assess goodness of fit of the predictive equation coefficients for

calculating passive moment contributions.   Normality of data was assessed using

Shapiro-Wilk tests, and the alpha level for all tests was set at 0.05.

3   Results

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

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

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

between LBP and NBP groups for age, gender, body mass (within gender), for any of the

gait parameterswalking speed or step length measured (P > 0.05).   There were no

significant differences (P > 0.05) in any of the physical activity parameters

measuredcalculated, which included walking (13.6 (SD 18.3) hours per week LBP, 9.9

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

4.1 (SD 5.6) hours NBP) and vigorous exercise (6.7 (SD 8.7) hours per week LBP, 5.1

(SD 3.4) hours NBP).

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

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

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

4   0.07) Nm/(kg*·m) for NBP. The adjusted r-squared values were 0.889 (SD 0.081) for LBP

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

6   between groups ($P > 0.05$).

7   ~~During early stance phase total hip moments were negative (hip extensor~~

8   ~~moments) and increasing to zero, becoming positive at 18% GC for LBP and 22% for~~

9   ~~NBP (figure 1). Throughout mid-to-late stance total hip moments were positive (hip flexor~~

10   ~~moments), with a peak (FL) at approximately 50% GC for both groups. From late stance~~

11   ~~to mid-swing, total hip moments were positive but decreasing towards zero.~~ There were

12   no statistically significant interaction effects in the total hip flexor (FL) or extensor moment

13   peaks (Ext1, Ext2) ~~or their timings~~ ($F(1.571,155)$ ~~(~~= 3.019, P = 0.065~~> 0.05~~, table 1) or

14   their timings ($F(1.599,158) = 0.368$, $P = 0.645$) (figure 2). There was no main effect of

15   group $F(1,99)=0.208$, $P = 0.649$). During the hip flexion component of the gait cycle,

16   there was no statistically significant interaction effect in total hip moments ($F(2.064,204)$

17   $= 0.823$, $P = 0.444$) (table 2). There was no main effect of group ($F(1,99) = 0.012$, $P =$

18   0.913).

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Hip Moments with 95% Confidence Intervals

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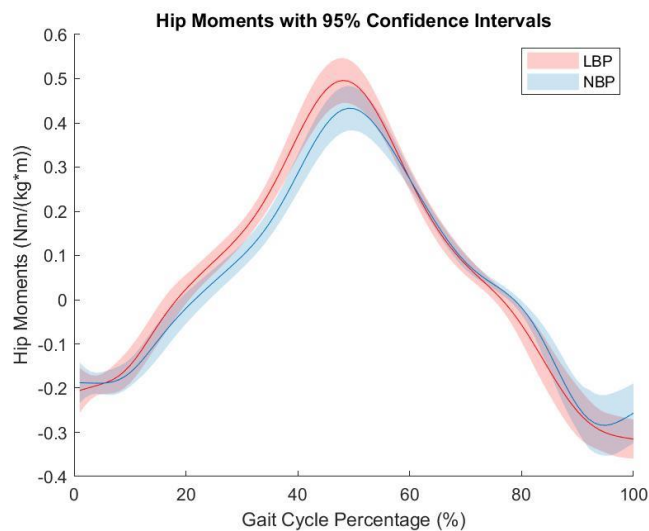
2 Figure 24. Mean total hip moments with 95% confidence intervals in LBP and NBP groups
3 during gait cycle.
4
5
6   For passive hip moments there was There was a significant interaction effect of

7 group and percentage of hip flexion for passive hip moments (F(1.751,173) = 9.316, P =<

8 0.000) and a statistically significant main effect of group (F(1,99) = 6.597, P = 0.012).

9 Passive hip moments decreased throughout hip flexion, demonstrating increased

10 extensor moments (figure 32), with no significant differences between groups at hip

11 neutral or 25% (P > 0.05). From 50-100% of hip flexion, passive moments were

12 significantly more negative in LBP subjects (P < 0.05), demonstrating greater extensor

13 moments. At maximum hip flexion, passive hip extensor moments were 46.6% and

14 39.4% of total hip moments, for LBP and NBP, respectively. Passive hip extensor and

15 total hip moment mean, standard deviation and results of the independent t-tests are

16 shown in table 2.

14

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

Moments (Nm/(kg*m))		Mean (SD)	Confidence Interval Lower	Upper	ANOVA *P*-value	Pairwise *P*-value
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		

15

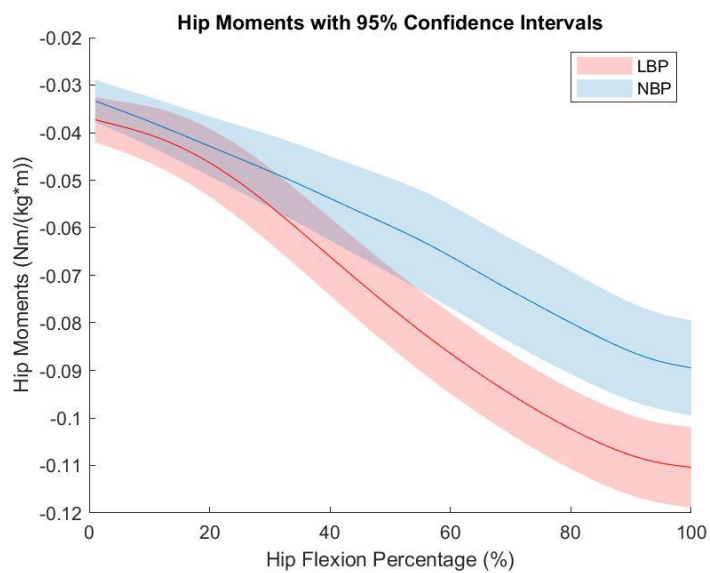**Hip Moments with 95% Confidence Intervals**

1

2   Figure 32. Mean passive hip extensor moments with 95% confidence intervals in LBP
3   and NBP groups during hip flexion from neutral to maximum hip flexion
4

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

			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	

(Row label: Hip Flexion Angle (%))

Total hip power was generated during the first 20% of the gait cycle, with an initial peak of power generation (H1) during early stance phase in both LBP and NBP (figure 3). 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 significantly earlier in LBP than NBP (8.2(4.5) %, 10.0(4.4) %, for LBP and NBP, respectively, $P = 0.049$), with no difference in power ($P > 0.05$. table 1, figure 4). The first peak of power generation (H1) occurred at

17

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 power ($P > 0.05$. table 1, figure 4). The hips then absorbed power, with a peak of power absorption (H2) during mid-stance. The peak of power absorption at H2 was significantly greater in LBP than NBP (-0.520 (SD 0.231) W/(kg*.m), -0.429 (SD 0.212) W/(kg*.m), for LBP and NBP, respectively, $P = 0.041$), with no difference in timings between groups ($P > 0.05$). Total hip power was positive from approximately 50% to 85% GC in both groups, as the hip flexors generated power from the end of stance phase through to the end of swing phase. Power generation peaked at toe-off and the initiation of swing phase, at approximately 60% GC (H3). The second peak of power generation at H3 was significantly greater in LBP than NBP (0.844 (SD 0.298) W/(kg*.m), 0.736 (SD 0.235) W/(kg*.m) for LBP and NBP, respectively, $P = 0.045$), with no difference in timings between groups ($F(1.526,151), = 0.863, P = 0.398$). Total hip power became negative from 90-100% of the GC in LBP, as the hip muscles absorbed power at the end of the swing phase and initial foot contact, where in NBP they were positive.
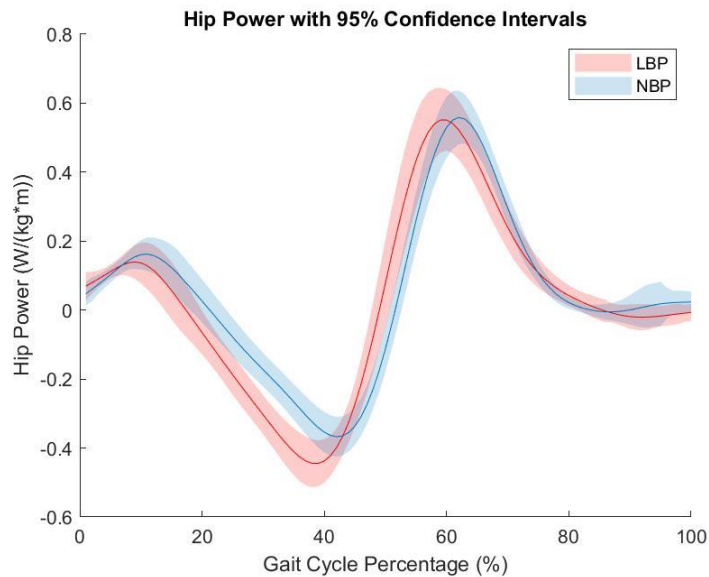
Hip Power with 95% Confidence Intervals

1

Figure 43. Mean total hip power with 95% confidence intervals in LBP and NBP groups
during gait cycle.

4

For power during the hip flexion component of gait (figure 5), ANOVA results

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

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

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

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

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

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

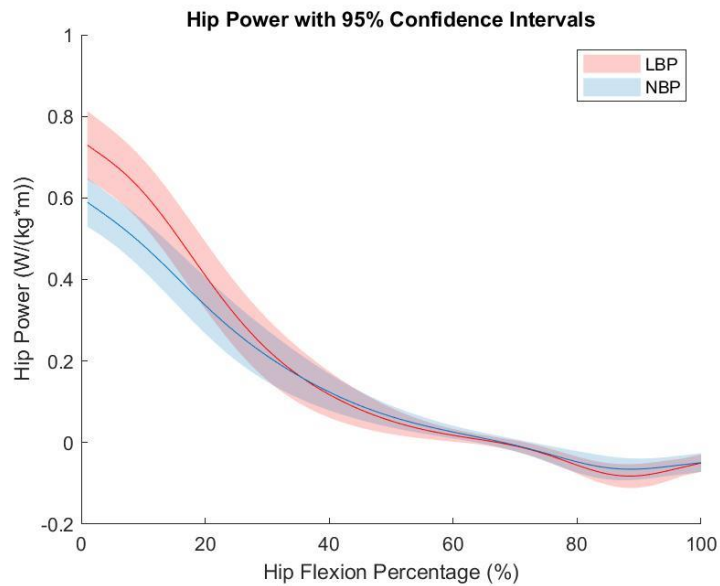t-tests for hip flexion are shown in table 3.

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2 Figure 54. Mean hip power with 95% confidence intervals in LBP and NBP groups during
3 hip flexion from neutral to maximum hip angle.
4
5 Table 3. Total hip power (W/(kg*m)) during hip flexion, from neutral to full hip flexion.
6 Statistically significant results are denoted with an asterisk(*).

			Power (W/(kg*m))				
				Confidence Interval		**ANOVA**	**Pairwise**
			Mean (S.D.)	**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		
Hip Flexion Angle (%)	**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		

7

20

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 during H2 (-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 (figure 6), there were no other 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

			Work Done (J/(kg*m))			
				Confidence Interval		
			Mean (S.D.)	Lower	Upper	*P-value*
Hip Flexion Angle (%)	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	

1

2   4   Discussion

3

4   The present study found no statistically significant differences in hip and knee angles

5   between groups. Values of passive hip extensor moments, and total hip moments, power

6   and work done for NBP subjects were similar to those reported elsewhere [24]. Other

7   investigators have reported alterations in hip and knee angles during walking in LBP

8   subjects compared with healthy controls [25-27]. The most likely cause of increased

9   passive moments in LBP is an increased passive resistance of the hip extensor and knee

10   flexor muscles, such as the bi-articular hamstrings muscles. However, a previous study

11   using the same subjects demonstrated no significant differences in passive moments

12   between groups. Because passive hip moments increase with hip angle in an exponential

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

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

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

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

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

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

19   passive moments occurred without differences in total moments.

20   **Peak hip flexor moments (FL) were greater in LBP than NBP, during the late**

21   **stance phase of gait. Hip flexor moments were also greater as the hip passed**

22   **through neutral during the early swing phase, with no differences in hip moments**

23   **during the remainder of hip flexion. Greater total hip flexor moments coincided**

~~with increased hip flexor power generation and mechanical work done in LBP compared with controls, as the hip passed through neutral in early swing phase. Increased hip moments may coincide with increased muscle activation, with increased power generation either as a pre-emptive strategy to overcome the increased passive resistance later in the swing phase, or to increase hip-pelvic-lumbar stability.~~

Power absorption and work done at H2 were significantly greater in LBP than NBP. The H2 power curve occurs during the stance phase of gait, as the hip flexors eccentrically contract to stabilise the pelvis and support the trunk, preventing backwards movement of the trunk relative to the pelvis. Power generation and work done at H3 were also greater in LBP than NBP, and occurs during the late stance (pre-swing) phase of gait, as the hip flexors contract concentrically to pull the leg forward and facilitate toe-off. This coincided with greater power generation and work in LBP than NBP as the hip moved through neutral. ~~Although there were no group differences in the H1 peak of power generation that follows initial contact in early stance, power generation occurred earlier in LBP than NBP.~~

The results of the present study demonstrate increased passive extensor moments in LBP, with no differences in total moments ~~during the same portion of hip flexion~~. Whether the passive moments contribute to total moments, thereby reducing the need for active contraction and promoting efficiency, or if values of total moments should be altered due to the influence of greater passive moments, is not currently understood [24]. Alternatively, altered joint angles, and therefore passive moments, later in the swing phase may follow the increased total power generation from the hip flexors during early

1 swing, directly following toe-off. In this case, increased passive moments may be due to

2 increased joint angles, resulting from greater momentum of the hip where active power

3 generation is greater in LBP with no differences in passive moments following toe-off.

4 Greater momentum of the swinging limb could cause greater hip flexion angles,

5 contributing to greater passive resistance, and therefore higher passive moments. The

6 lack of difference in total moments may reflect reduced active components late in swing,

7 or be due to total moments being miscalculated due to lack of account of passive

8 influences in inverse dynamic modelling. Despite our lack of understanding regarding the

9 mechanisms and interactions of the passive and active components, it is clear that the

10 passive component is considerable, and should not be overlooked in future models of

11 dynamic movements.

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

13 3.1)) compared with other studies, VAS scores were considered moderate (5.06 (SD

14 2.27)), and similar to those reported elsewhere [28-35]. Although self-reported pain is a

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

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

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

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

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

20 with healthy controls.

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

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

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

24

and moments during movement, sufficient to influence passive components without affecting total moment. There may be further interactions between LBP, passive moments, total power and total work done, which affect accelerations and energy efficiency for movement. From a clinical perspective, an assessment of passive biomechanics will be useful to help understand which tissue types are influenced by LBP, on a case-by-case basis. Findings suggestive of structural adaptations will necessitate a muscle-targeted therapy, such as stretching or strengthening exercise. Conversely, if joint kinematics are altered, this indicates the active component of the muscular system may have been affected, necessitating a therapy that targets neuromuscular improvements, such as muscle activation timing and magnitude, and the specific fibres recruited during movement.

A limitation of the current study was the measurement of hip passive moments during hip flexion only. In previous studies [24, 36, 37], researchers averaged hip moments during flexion and extension, and reported only a small underestimation of moments during flexion, and overestimation during extension. However, in the present study it was considered more appropriate to determine accurate values of passive hip extensor moments, for integration into the hip flexion model during walking, rather than over- or underestimating moments. Future studies would benefit from direct measurement of both passive hip extensor and hip flexor moments to model the complete gait cycle accurately in the sagittal plane.

Hip extensor moments were 46.6% and 39.4% of total hip moments for LBP and NBP, respectively, at maximum hip flexion angle. This finding demonstrates that passive structures may interact with active components, have a considerable influence on total

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

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

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

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

5 further investigation.

6

7 5   Conclusions

8

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

10 moments and total ~~hip moments,~~ power and work done during walking, compared with

11 healthy controls. Although it is not possible to extrapolate cause and effect relationships,

12 rehabilitation programmes for LBP patients should differentiate between the active,

13 neuromuscular components of movement, and the passive components. Biomechanical

14 models should include individual measurements of passive joint moments. The approach

15 used in the present study may be a useful measurement model for clinicians assessing

16 low back pain.

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