University of Massachusetts Amherst

From the Selected Works of Joseph Hamill

December, 2008

In Vivo Lumbo-Sacral Forces and Moments during Constant Speed Running at Different Stride Lengths

Joseph Seay

W. Scott Selbie

Joseph Hamill, *University of Massachusetts - Amherst*

Available at: https://works.bepress.com/joseph_hamill/10/

Journal of Sports Sciences

Publication details, including instructions for authors and subscription information:

<http://www.tandfonline.com/loi/rjsp20>

In vivo lumbo-sacral forces and moments during constant speed running at different stride lengths

Joseph Seay^a, W. Scott Selbie^b & Joseph Hamill^a

^a Department of Kinesiology, University of Massachusetts Amherst, Massachusetts, Amherst, USA

^b C-Motion, Inc., Germantown, Maryland, USA

Published online: 27 Nov 2008.

To cite this article: Joseph Seay, W. Scott Selbie & Joseph Hamill (2008) In vivo lumbo-sacral forces and moments during constant speed running at different stride lengths, Journal of Sports Sciences, 26:14, 1519-1529, DOI: [10.1080/02640410802298235](https://doi.org/10.1080/02640410802298235)

To link to this article: <http://dx.doi.org/10.1080/02640410802298235>

PLEASE SCROLL DOWN FOR ARTICLE

Taylor & Francis makes every effort to ensure the accuracy of all the information (the "Content") contained in the publications on our platform. However, Taylor & Francis, our agents, and our licensors make no representations or warranties whatsoever as to the accuracy, completeness, or suitability for any purpose of the Content. Any opinions and views expressed in this publication are the opinions and views of the authors, and are not the views of or endorsed by Taylor & Francis. The accuracy of the Content should not be relied upon and should be independently verified with primary sources of information. Taylor and Francis shall not be liable for any losses, actions, claims, proceedings, demands, costs, expenses, damages, and other liabilities whatsoever or howsoever caused arising directly or indirectly in connection with, in relation to or arising out of the use of the Content.

This article may be used for research, teaching, and private study purposes. Any substantial or systematic reproduction, redistribution, reselling, loan, sub-licensing, systematic supply, or distribution in any form to anyone is expressly forbidden. Terms & Conditions of access and use can be found at <http://www.tandfonline.com/page/terms-and-conditions>

***In vivo* lumbo-sacral forces and moments during constant speed running at different stride lengths**

JOSEPH SEAY¹, W. SCOTT SELBIE², & JOSEPH HAMILL¹

¹Department of Kinesiology, University of Massachusetts Amherst, Massachusetts, Amherst, USA and

²C-Motion, Inc., Germantown, Maryland, USA

(Accepted 24 June 2008)

Abstract

The aim of this study was to introduce a Newton–Euler inverse dynamics model that included reaction force and moment estimation at the lumbo-sacral (L5–S1) and thoraco-lumbar (T12–L1) joints. Data were collected while participants ran over ground at $3.8 \text{ m} \cdot \text{s}^{-1}$ at three different stride lengths: preferred stride length, 20% greater than preferred, and 20% less than preferred. Inputs to the model were ground reaction forces, bilateral lower extremity and pelvis kinematics and inertial parameters, kinematics of the lumbar spine and thorax and inertial parameters of the lumbar segment. Repeated measures ANOVA were performed on the lower extremity sagittal kinematics and kinetics, including L5–S1 and T12–L1 three-dimensional joint angles, reaction forces and moments at touchdown and peak values during impact phase across the three stride conditions. Results indicated that L5–S1 and T12–L1 vertical reaction forces at touchdown and during the impact portion of the support phase increased significantly as stride length increased ($P < 0.001$), as did peak sagittal L5–S1 moments during impact ($P = 0.018$). Additionally, the transverse T12–L1 joint moment increased as running speed increased ($P = 0.006$). We concluded from our findings that our model was sensitive to our perturbations in healthy runners, and may prove useful in future mechanistic studies of L5–S1 mechanics.

Keywords: *Inverse dynamics, biomechanics, low back, running*

Introduction

Low back pain is a global issue (Andersson, 1997). Approximately 10% of recreational runners experience low back pain within their first year of training (Jacobs & Bernsen, 1986; Taunton *et al.*, 2003), and injuries in the region of the pelvis and low back can require extended bouts of rehabilitation before function is restored and pain resolved (Fields, Kramer, & Delaney, 1990; Geraci, 1998).

Although clinical locomotion studies have revealed kinematic asymmetries between people with and without lower back pain, such as during increases in gait speed (Taylor, Evans, & Goldie, 2003; Vogt & Banzer, 1999), and during level and uphill walking (Vogt, Pfeifer, Portscher, & Banzer, 2001), there are few studies that examined kinetic differences between individuals with and without low back pain. Although musculoskeletal models have provided insight into the strain on the soft tissue surrounding the spine during various activities in healthy individuals (Callaghan, Patla, & McGill, 1999; Khoo, Goh, & Bose, 1995), these

models have not yet developed to the point where they are able to discern mechanistic differences between individuals with and without low back pain. Because ground reaction forces are dampened as they travel through the body, most studies that model the lower back have computed forces or torques from the cranial aspect, using “top down” approaches that do not factor in the contribution of transient forces due to impact during the gait cycle. These transient forces that result from the foot-ground collision are important in the study of locomotion because they typically make a large contribution to the joint reaction forces and joint moments during early stance, and have been implicated in lower extremity running injuries (e.g. Callaghan *et al.*, 1999; Voloshin & Wosk, 1982).

Many studies that estimate joint forces and moments use a Newton–Euler modeling technique. This type of model has been useful to clinicians and researchers in evaluating differences in lower extremity gait mechanics between injured and non-injured runners (Milner, Ferber, Pollard, Hamill, & Davis, 2006), and in older individuals with and

without knee osteoarthritis (Baliunas *et al.*, 2002; Messier *et al.*, 2005).

Newton-Euler models may elucidate the effect of kinetic differences at the low back caused by the foot-ground interaction during running. Although these link-segment models have their limitations (see Winter, 1990), with careful interpretation they can lend insight into how the lower extremity system is adapting to change. For example, Derrick, Hamill, and Caldwell (1998) documented that lower extremity sagittal joint moments and powers increased in response to increasing stride length while maintaining a constant running speed of $3.8 \text{ m} \cdot \text{s}^{-1}$.

The purpose of this study was to develop a Newton-Euler model that included reaction force and moment estimation at the lumbo-sacral (L5-S1) and thoraco-lumbar (T12-L1) joint during running. The L5-S1 joint was chosen because anatomically it is the point at which the axial skeleton meets the appendicular skeleton, and is the site where over 85–95% of disc herniations occur (Armstrong, 1965). We tested our model on healthy active individuals by altering stride length while running at a constant speed. Based on previous findings by Derrick *et al.* (1998), we hypothesised that lumbo-sacral reaction forces and moments would increase as stride length increased while maintaining a constant speed.

Methods

Participants

Data were collected from ten healthy, young adults ranging in age from 22 to 32 years (mean age 26.2 years, $s = 2.9$), in height from 1.58 to 1.85 m (mean height 1.72 m, $s = 0.14$) and in body mass from 56.8 to 85.0 kg (mean 66.2 kg, $s = 10.2$). All participants were physically active, free of lower extremity injury, had no evidence of current low back pain, and no history of low back pain. All participants signed an informed consent form that was approved by the University of Massachusetts Amherst Human Use Review Committee, and completed a Physical Activity Readiness questionnaire (PAR-Q). A positive response to any question on the PAR-Q resulted in exclusion from the study.

Experimental set-up

Kinematic data were collected using eight MC240 cameras (Qualysis, Gothenburg, Sweden) sampling at 240 frames per second, and ground reaction data were collected at 960 Hz on one six-channel AMTI force platform (Advanced Mechanical Technologies, Watertown, MA, USA). The eight cameras were placed in such a way that the force platform was in the centre of the data collection volume, and both

were interfaced to an IBM-compatible computer. Data were collected using Qualysis Track Manager (QTM) software.

Modelling the lumbar spine

To create a Newton-Euler model that estimated reaction force and moment parameters at the lumbo-sacral (L5-S1) and thoraco-lumbar (T12-L1) joints, a bilateral lower extremity Newton-Euler model was extended to include lumbar and thoracic segments. The lumbar segment was defined such that the lumbar spine and surrounding soft tissue were modelled as a single rigid segment.

Clusters of passive reflective markers attached to rigid plastic shells were secured bilaterally on the thigh, shank, and heel (Manal, McClay, Richards, Galinat, & Stanhope, 2002). Markers were placed on bilateral trochanters, and on medial and lateral knee, ankle, and metatarsal heads during a standing calibration trial (Milner *et al.*, 2006) (Figure 1).

To track the pelvis, passive reflective markers were attached over the L5-S1 intervertebral joint space (L5S1 marker) and anterior superior iliac spines (ASIS). To define the local segment coordinate systems for pelvis, markers were placed on the following bilateral bony landmarks: ASIS, posterior superior iliac spine, greater trochanter (GT), and iliac crest (ILCR, directly superior to the trochanters).

Elastikon elastic tape (Johnson & Johnson, New Brunswick, NJ, USA) was wrapped around the lumbar region from just above the iliac crest to the ribcage at the level of the T12-L1 joint space to minimise soft tissue artefact. To track the motion of the lumbar spine, individual markers were placed over the Elastikon at the level of the T12-L1 joint space, and midway between the T12L1 marker and the L5S1 marker. Four markers were placed over the Elastikon on the lumbar region on either side of the midline markers with a minimum distance of 4 cm between all markers (Figure 1).

Local segment coordinate systems for the lumbar segment were defined using calibration markers placed bilaterally on the iliac crest and bilaterally on the ribcage at the level of the T12-L1 joint space immediately superior to the iliac crest markers (Figure 1). The local coordinate system for the thoracic segment was defined using bilateral ribcage and acromioclavicular (AC) markers (Figure 1), and the motion of the thorax segment was tracked using the T12L1 marker and bilateral AC markers.

Two segmental coordinate systems were adopted for the pelvis. A *technical coordinate system* was adopted such that the segment end-points of the model could meet (i.e. superior end of pelvis and inferior end of lumbar segment) in order to calculate kinetics at the L5-S1 joint, which was represented as

Figure 1. (A) Visual representation of the marker set used in the study. (B) Picture of marker set used to track the motion of the lumbar spine during running trials.

a virtual point created anterior to the L5S1 marker (Figure 2) on the superior posterior aspect of the pelvis. Specifically, the virtual point was created by creating a point 5% along the length of a virtual line that ran from the L5S1 marker to the bisector of the two ASIS markers. This point defined the L5-S1 joint centre in the technical coordinate system, based on the depth of the pelvis, in order to approximate the location of the anatomic L5-S1 inter-vertebral joint (Khoo *et al.*, 1995).

An additional segmental coordinate system, the *anatomical coordinate system*, was created to resolve kinematic and kinetic results into a more clinically recognisable frame of reference. The anatomical coordinate system was created by aligning the standing pelvis segment with the ILCR marker, which was directly superior to the GT marker. Even though this adjustment was performed primarily in the sagittal plane (Figure 2), deviations from this posture during locomotion were more clinically recognisable in pelvic tilt (sagittal plane), pelvic obliquity (frontal plane) and axial rotation (transverse plane).

Protocol

During a single visit to the biomechanics laboratory, participants were measured for body mass and height after signing informed consent forms that were approved by the University of Massachusetts Amherst Human Use Review Committee. Markers were then placed on participants as described earlier. With all calibration and motion markers in place, the participants stood near the centre of the force

Figure 2. Visual representation, from the sagittal view, of the technical coordinate system and anatomical coordinate system (see text for details). The solid lines represent the manner in which the technical coordinate system was resolved, and the $\oplus$ where the solid lines intersect represents the virtual point about which the lumbo-sacral forces were realised in the technical coordinate system. Note that the virtual L5-S1 point is located along the line (dotted line) that connects the L5-S1 joint space (L5-S1) and a virtual point midway between the left and right ASIS markers (m). The markers were placed on the subject such that the ILCR marker was placed directly superior to the GT marker, so that the pelvis was aligned vertically during the standing “anatomical neutral” pose. This allowed the anatomical coordinate system to be resolved to a more vertical alignment to reference standing posture (dashed line).

platform, facing in the direction of progression (+Y), with arms folded across the chest, and with feet hip width apart and parallel. Data were collected, and this static posture served as a template from which segmental coordinate systems were calculated. Once the calibration trial was collected, the calibration markers were removed.

The remainder of the protocol was executed according to Derrick *et al.* (1998). Participants were asked to run at a constant speed of $3.83 \text{ m} \cdot \text{s}^{-1}$ ($s=2.5\%$) along a 15-m runway. Speed was determined using two photo-electric cells located in the filming volume 5 m apart. Once participants were able to run at $3.83 \text{ m} \cdot \text{s}^{-1}$ ($s=0.09$) consistently, average stride length over five trials was calculated to generate the preferred stride length (PSL). Participants then performed running trials in three stride length manipulations (conditions): PSL, PSL-20% (20% shorter than PSL) and PSL+20% (20% longer than PSL). Touchdown sites for each condition were physically marked on the floor through the entire data capture volume, and participants were instructed to adjust their stride length accordingly. After ample practice to become accustomed to each stride condition, participants were required to complete at least five successful trials per right and left limbs for each condition. A successful trial was defined as the subject maintaining a constant speed of $3.83 \text{ m} \cdot \text{s}^{-1}$ throughout the trial and landing on the force platform. Stride length conditions were presented in balanced order, and participants were allowed to rest between conditions to prevent fatigue from influencing the trials. All participants wore the same model neutral running shoes (Saucony Jazz, The Stride Rite Corporation, Lexington, MA, USA) to control for potential effects of footwear.

Data reduction

Visual3D motion analysis software (C-Motion Inc., Germantown, MD, USA) was used to filter kinematic data and calculate three-dimensional, bilateral segmental pose (position and orientation) for foot, leg, thigh, pelvis and lumbar segments. Kinematic data were filtered using a low pass, fourth-order Butterworth filter at 10 Hz (Winter, 1990).

Cardan-Euler angles were calculated using an Xyz (flexion-extension, abduction-adduction, internal-external rotation) rotation order to describe the motions of the distal segment relative to the proximal segment for bilateral hip, knee, and ankle, as well as L5-S1 and T12-L1 joints. All joint angles, moments, and reaction forces, were calculated using a right-hand rule convention, and were expressed relative to the proximal segment. L5-S1 reaction forces and

moments were described with respect to the lumbar segment, and T12-L1 kinetic profiles were described relative to the thoracic segment.

Lower extremity segment masses were computed based on Dempster (1955). Pelvis moments of inertia were based on assuming that the pelvis is an elliptical cylinder with length from the iliac crest to GTs and a diameter equal to the distance between the GTs and a depth equal to the distance from mid-ASIS to L5S1. The lower extremity moments of inertia were computed by representing each segment as a frustum of a right cone with proximal and distal diameters equal to the segment diameters. Lumbar segment anthropometry was computed according to Pearsall, Reid, and Livingston (1996), and lumbar moments of inertia were based on the assumption that the lumbar spine segment is a cylinder that runs from the ribcage markers to the iliac crest markers, with three-dimensional moments of inertia as described by Pearsall *et al.* (1996). The thorax segment was defined as a cylinder that ran from the AC markers to the ribcage markers at the T12-L1 level.

The resulting signals were time-normalised to 100% of stance using cubic spline interpolation.

Statistical analysis

A one-way analysis of variance (ANOVA) with repeated measures was used to compare peak reaction force and moment profiles between stride length manipulations at touchdown and during the impact portion of the support phase. Peak values were averaged across all trials within participants, then averaged for each stride length manipulation. Impact phase (impact) was defined as the first 50 ms after touchdown (Nigg, 1997). An *a priori* level of $\alpha=0.05$ was set for the comparison between stride length manipulations, and Tukey *post hoc* analyses were used to determine significance between pairs of stride lengths where significance was observed in the ANOVA.

Results

Kinematics and kinetics around the lumbar segment

The mean right support limb histories of L5-S1 angles and moments for one exemplar subject over all conditions are presented in Figure 3, and T12-L1 angles and moments are presented in Figure 4. All data are presented with respect to a right side support limb.

No significant differences were observed between sagittal, frontal or transverse L5-S1 joint angles at touchdown or during impact (Table I). L5-S1 axial reaction forces increased significantly at touchdown ($P < 0.001$) and during impact ($P < 0.001$) as stride length increased. *Post hoc* analyses revealed that axial

Figure 3. Mean L5-S1 joint angle and moment profiles during support phase during all three stride conditions for one exemplar subject. FLEX, flexion; EXT, extension; CCW (CW), pelvis segment oriented counterclockwise (clockwise) to lumbar segment.

reaction force was significantly greater at touchdown and during impact for PSL + 20% compared with both PSL (for touchdown and impact, adj $P > 0.001$) and PSL - 20% (for touchdown, adj $P = 0.004$; during impact, adj $P = 0.008$) (Table I). L5-S1 sagittal joint moments also increased significantly during impact as stride length increased ($P = 0.018$), and *post hoc* analyses revealed that sagittal L5-S1 moment was significantly greater for the longest stride length (PSL + 20%) compared with both PSL (adj $P = 0.038$) and PSL - 20% (adj $P = 0.031$) (Table I). No other significant differences were observed between any other L5-S1 parameters at touchdown or during impact.

Significant main effects were observed for peak T12-L1 range of motion in all the three planes of motion as the stride length increased (Table II), with the most significant changes occurring in the transverse plane ($P < 0.001$). *Post hoc* analyses revealed that the range of motion for the longest

stride length (PSL + 20%) was significantly greater than PSL - 20% in the sagittal and frontal planes (Table II). In the transverse plane, peak range of motion for T12-L1 was significantly greater for PSL + 20% than both shorter stride lengths.

T12-L1 axial reaction forces increased significantly at touchdown ($P < 0.001$) and during impact ($P = 0.001$) as stride length increased (Table II). *Post hoc* analyses revealed that axial reaction force was significantly greater at touchdown and during impact for PSL + 20% compared with both PSL (touchdown, adj $P = 0.005$; impact, adj $P = 0.026$) and PSL - 20% (touchdown and impact, adj $P < 0.001$) (Table II). T12-L1 antero-posterior reaction forces also increased at touchdown as stride length increased ($P = 0.001$), with AP reaction force significantly greater at touchdown for PSL + 20% compared with both PSL - 20% (adj $P < 0.001$) and PSL (adj $P = 0.005$) (Table II). T12-L1 transverse joint moments also increased significantly at

Figure 4. Mean T12-L1 joint angle and moment profiles during support phase during all three stride conditions for one exemplar subject. FLEX, flexion; EXT, extension; CCW (CW), pelvis segment oriented counterclockwise (clockwise) to lumbar segment.

touchdown ($P=0.039$) and during impact ($P=0.006$) as stride length increased. *Post hoc* analyses revealed that transverse L5-S1 moment was significantly greater for PSL + 20% as compared with PSL - 20% at touchdown (adj $P=0.036$), and significantly greater during impact for PSL + 20% as compared with both PSL (adj $P=0.038$) and PSL - 20% (adj $P=0.031$) (Table I). No other significant differences were observed between any other T12-L1 parameters at touchdown or during impact.

Bilateral vertical joint reaction force profiles for all lower extremity, L5-S1 and T12-L1 joints from one exemplar subject are presented in Figure 5. Both right leg (support limb) and left leg (swing limb) joint reaction forces are included to demonstrate the bilateral contribution to the lumbar reaction force pattern. Both L5-S1 and T12-L1 vertical joint reaction curves demonstrated a small compression

peak during the first 20% of support, followed by a larger peak at around 45% of support.

Kinematics and kinetics of the lower extremity

Mean sagittal joint angles and moments for the lower extremity joints are presented in Table III. Ankle plantar flexion angle increased significantly at touchdown as stride length increased, as did peak hip extension angle during impact. Hip and knee extension moments also increased significantly as stride length increased, both at touchdown and during impact.

Discussion

The purpose of this study was to introduce a Newton-Euler model that included joint reaction force and joint moment estimation at the L5-S1 and T12-L1

Table I. Ensemble means over all participants and all trials for lumbar (L5-S1) joint kinematics, forces and moments.

L5-S1 parameter	Condition	Touchdown			Impact phase		
		Mean	<i>s</i>	<i>P-value</i>	Mean	<i>s</i>	<i>P-value</i>
Joint angle (°)							
Sagittal	PSL − 20%	−0.7	6.1	0.073	3.8	5.6	0.165
	PSL	1.1	4.4		5.3	4.5	
	PSL + 20%	−3.2	6.7		1.9	6.3	
Frontal	PSL − 20%	−1.7	2.1	0.107	−1.5	2.1	0.104
	PSL	−2.9	1.7		−2.7	1.7	
	PSL + 20%	−3.1	2.8		−2.9	2.7	
Transverse	PSL − 20%	5.1	0.8	0.869	2.9	1.0	0.762
	PSL	5.6	2.2		3.7	1.7	
	PSL + 20%	5.3	4.2		3.3	4.4	
Joint moment (Nm · kg ^{−1})							
Sagittal	PSL − 20%	0.6	0.5	0.413	−2.1	0.5	0.018*
	PSL	0.3	0.3		−2.1	0.7	
	PSL + 20%	0.6	0.4		−2.7	1.0	
Frontal	PSL − 20%	0.2	0.3	0.874	−0.2	0.3	0.893
	PSL	0.2	0.2		−0.1	0.3	
	PSL + 20%	0.2	0.2		−0.1	0.4	
Transverse	PSL − 20%	−0.3	0.2	0.401	−0.7	0.2	0.577
	PSL	−0.2	0.2		−0.6	0.1	
	PSL + 20%	−0.2	0.1		−0.6	0.2	
Reaction force (BW)							
Vertical	PSL − 20%	0.03	0.06	< 0.001*	−0.7	0.2	< 0.001*
	PSL	0.05	0.08		−0.8	0.3	
	PSL + 20%	0.14	0.07		−1.0	0.3	
Anterior/posterior	PSL − 20%	0.04	0.09	0.069	−0.06	0.06	0.337
	PSL	−0.01	0.06		−0.05	0.03	
	PSL + 20%	−0.02	0.03		−0.03	0.03	
Medial/lateral	PSL − 20%	0.03	0.05	0.177	−0.04	0.06	0.959
	PSL	0.00	0.04		−0.04	0.03	
	PSL + 20%	−0.02	0.02		−0.03	0.02	

Note: All kinetic data are normalised to body weight (BW).

*Statistical significance, set at the $P=0.05$ level. Similar numbers to the right of the P -value indicate no significant *post hoc* differences between condition means.

joints, and to test this model using a perturbation that would elicit a predictable response in healthy runners. We found the model to be sensitive to stride length alterations during constant speed running. Specifically, as stride length increased, the impact of the foot-ground collision increased. As a result, we observed increased L5-S1 vertical reaction forces as stride length increased at touchdown and during impact, and increased sagittal L5-S1 moments during impact. Results also indicated that vertical reaction force at T12-L1 increased as stride length increased at touchdown and during impact, and antero-posterior reaction forces also increased at touchdown. T12-L1 transverse moment also increased at touchdown and during impact as stride length increased.

The patterns and ranges of our lumbar ranges of motion for our group running at PSL were consistent with the previous literature that has reported lumbosacral (L5-S1) ranges of motion during running. Whittle, Levine, and Pharo (2000) reported a range of 12.1° for lumbar flexion/extension during running at

$2.9 \text{ m} \cdot \text{s}^{-1}$ and Schache, Blanch, Rath, Wrigley, and Bennell (2002) reported lumbar ranges of motion of 13.3° in flexion/extension, 18.5° in lateral bending and 23.0° in axial rotation over the entire gait cycle for healthy individuals running at $4.0 \text{ m} \cdot \text{s}^{-1}$. Our group means were consistent with these findings for flexion/extension (mean 6.59° , $s=2.08$), lateral bending (mean, 11.54° , $s=3.32$) and axial rotation (mean 12.41° , $s=2.73$) during support phase for healthy individuals running their PSLs at $3.8 \text{ m} \cdot \text{s}^{-1}$. The main differences between our results and those of Whittle *et al.* (2000) and Schache *et al.* (2002) stem primarily from the fact that these investigators are reporting ranges of motion over the entire gait cycle, whereas in the current investigation we report only the support phase kinematics. The differences could also be a function of tracking marker setups used to define the lumbar spine. The investigators cited defined the lumbar segment using rigid marker triads, one of which was on a rod that extended away from the body, which could potentially increase the three-dimen-

Table II. Ensemble means over all participants and all trials for T12-L1 joint kinematics, reaction forces and moments.

T12-L1 parameter	Condition	Touchdown			Impact phase				
		Mean	<i>s</i>	<i>P-value</i>	Mean	<i>s</i>	<i>P-value</i>		
Joint angle (°)									
Sagittal	PSL − 20%	−17.8	11.5	0.047*	¹ ²	−17.5	11.8	0.041*	¹ ¹
	PSL	−15.8	9.8			−15.2	9.8		
Frontal	PSL + 20%	−9.6	14.7	0.043*	¹ ² ²	−8.8	14.5	0.022*	¹ ¹ ¹
	PSL − 20%	−5.3	2.7			−4.3	3.1		
	PSL	−6.2	3.0			−5.3	3.5		
Transverse	PSL + 20%	−7.8	4.4	<.001*	¹ ¹ ² ²	−6.9	4.1	<0.001*	¹ ¹ ¹ ²
	PSL − 20%	5.3	4.3			1.0	3.9		
	PSL	7.4	5.3			2.1	3.9		
	PSL + 20%	11.5	5.7			5.1	5.7		
Joint moment (Nm · kg ^{−1})									
Sagittal	PSL − 20%	0.5	0.6	0.481		−1.8	0.5	0.099	
	PSL	0.3	0.3			−1.7	0.6		
	PSL + 20%	0.4	0.4			−2.2	0.8		
Frontal	PSL − 20%	0.2	0.4	0.875		−0.5	0.4	0.531	
	PSL	0.2	0.3			−0.4	0.3		
	PSL + 20%	0.2	0.2			−0.5	0.5		
Transverse	PSL − 20%	−0.3	0.1	0.039*	¹ ² ²	−0.6	0.1	0.006*	¹ ¹ ²
	PSL	−0.3	0.1			−0.6	0.1		
	PSL + 20%	−0.2	0.1			−0.8	0.2		
Reaction force (BW)									
Vertical	PSL − 20%	0.0	0.08	<0.001*	¹ ¹ ²	−0.6	0.2	0.001*	¹ ¹ ²
	PSL	0.1	0.09			−0.6	0.3		
	PSL + 20%	0.2	0.07			−0.8	0.3		
Anterior/posterior	PSL − 20%	0.0	0.1	<0.001*	¹ ¹ ²	−0.1	0.08	0.069	
	PSL	−0.1	0.08			−0.1	0.06		
	PSL + 20%	−0.1	0.04			−0.1	0.02		
Medial/lateral	PSL − 20%	0.0	0.05	0.461		−0.1	0.04	0.215	
	PSL	0.0	0.05			0.0	0.07		
	PSL + 20%	0.0	0.06			0.1	0.06		

Note: All kinetic data are normalised to body weight (BW).

*Statistical significance, set at the $P=0.05$ level. Similar numbers to the right of the P -value indicate no significant *post hoc* differences between condition means.

sional range of motion during running due to the inertial properties of the rod, no matter how light. We used seven individual markers that were spread across the lumbar spine (Figure 1), and tracked the motion of our lumbar segment by modelling those markers as a cluster.

As hypothesised, we observed increases in peak reaction forces at touchdown and during impact at both L5-S1 (Table I) and T12-L1 joints as stride length increased (Table II). Qualitatively, we observed the vertical reaction force was notably diminished at the L5-S1 joint when compared with the support limb hip joint (Figure 5). The lumbar reaction force was most likely dispersed as an additive function of the swing leg mechanics (i.e. because the swing leg is not in contact with the ground), and therefore attenuated the force realised at L5-S1. The reaction forces generated by the swing limb, although much smaller, were generated in the opposite direction of the reaction forces being calculated for the stance limb (i.e. a 'tension' reaction force [positive vertical joint

reaction force, see Figure 5]) as opposed to a 'compression' reaction force [negative vertical joint reaction force]). This 'tensile' vertical reaction force also contributes to the attenuation of the 'compressive' vertical reaction force realised at L5-S1.

Significant differences were observed with an increased stride length in the T12-L1 kinematics as compared with the L5-S1 kinematics. Although peak T12-L1 sagittal plane (thoraco-lumbar flexion) and frontal plane (thoraco-lumbar obliquity) angles increased as stride length increased during a constant velocity, the largest differences were observed in the transverse plane (thoraco-lumbar axial rotation). These observations are consistent with the anatomy of the articulations on either end of the lumbar spine, in which the T12-L1 joint is most influenced by the inertia of the ribcage and arms in the transverse plane during running. Conversely, the L5-S1 joint is supported more by the musculature surrounding the pelvis and lumbar spine during running. The observed differences between stride length conditions,

Figure 5. Mean (solid lines) $\pm$ s (broken lines) vertical lower extremity and lumbar joint reaction force profiles during support phase during PSL condition for one exemplar subject running at $3.8 \text{ m} \cdot \text{s}^{-1}$. All forces are normalised to body weight.

Table III. Ensemble means over all participants and all trials for sagittal lower extremity joint angle and moments.

Lower extremity parameter	Condition	Touchdown				Impact phase				
		Mean	<i>s</i>	<i>P-value</i>		Mean	<i>s</i>	<i>P-value</i>		
Sagittal joint angle (°)										
Ankle	PSL − 20%	18.9	4.8		¹	8.6	3.6			
	PSL	20.1	5.6	0.043*	¹	6.8	2.6	0.079		
	PSL + 20%	21.3	3.3			5.1	4.1			
Knee	PSL − 20%	−14.5	7.0			−31.2	9.5			
	PSL	−13.3	6.8	0.353		−32.5	9.2	0.090		
	PSL + 20%	−16.1	6.0			−37.4	5.0			
Hip	PSL − 20%	31.5	6.5e			32.0	6.4		¹	
	PSL	32.1	4.7	0.051		33.1	4.7	0.023*	¹	²
	PSL + 20%	35.9	4.0			37.0	3.7			²
Sagittal joint moment (Nm · kg ^{−1})										
Ankle	PSL − 20%	0.0	0.04			−0.7	0.3			
	PSL	0.0	0.04	0.272		−0.7	0.3	0.202		
	PSL + 20%	0.0	0.09			−0.9	0.4			
Knee	PSL − 20%	−0.3	0.1		¹	0.9	0.6		¹	
	PSL	−0.2	0.2	0.015*	¹	1.3	0.7	<0.001*	²	
	PSL + 20%	−0.1	0.2			1.6	0.7		³	
Hip	PSL − 20%	−0.6	0.3		¹	−2.5	0.6		¹	
	PSL	−0.6	0.4	0.005*	¹	−2.6	0.8	0.010*	¹	²
	PSL + 20%	−0.1	0.3		²	−3.1	0.9			²

Note: All joint angles are normalised to a standing calibration trial, and kinetic data are normalised to body weight.

*Statistical significance, set at the $P=0.05$ level. Similar numbers to the right of the P -value indicate no significant *post hoc* differences between condition means.

namely increases in vertical reaction forces and sagittal moments, were primarily due to the increased impact that accompanied longer stride lengths.

We acknowledge the limitations to our model of the lumbar spine. Unlike the lower extremity and pelvis, there is almost no physical association between the soft tissue (including viscera) and the lumbar spine. In addition to this limitation, the parameters from Pearsall *et al.* (1996) were estimated from CT scans during which the participant was lying supine. Pearsall notes, and we agree, that spinal column mass distributions may be different when a posture other than supine is assumed. However, regardless of these limitations, we still observed increases in reaction forces and moments around the lumbar spine as stride length increased.

When compared with the musculo-skeletal modelling literature regarding estimation of lumbar forces, our results are quite different. For instance, vertical lumbo-sacral loads in the range of 1.45 and 2.07 times bodyweight (Khoo *et al.*, 1995) and 1.0–2.5 times bodyweight (Callaghan *et al.*, 1999) have been reported vertical for healthy participants at preferred walking speeds. In general, these values are greater in magnitude than measured vertical ground reaction forces for walking, which tend to be about 1.1–1.2 times bodyweight (Winter, 1990). By contrast, for an exemplar subject, who produced a peak ground reaction force of about 2.3 times bodyweight while running at $3.8 \text{ m} \cdot \text{s}^{-1}$, our model indicated

that the peak joint reaction force estimated at our lumbo-sacral joint did not exceed 1.5 times bodyweight. The peak reaction force over the entire stance phase in our model was closer to 1.3 times bodyweight while running at PSL. The main source of this difference stems from the fact that the joint reaction force is the minimum force of interaction possible between the segments; all muscular inputs and tissue constraints result in higher forces. Although we agree that muscular contraction plays a role in lumbar activity, the link-segment model used here is designed to estimate the lumbar reaction forces. In our study, this model was used effectively to monitor changes in lower extremity and L5-S1 and T12-L1 joint kinetics (i.e. the lower extremity and lumbar “system”). We observed changes in this system as we applied a known perturbation to it (i.e. by increasing stride length while maintaining a constant speed), and conclude from our results that this model was sensitive enough to differentiate between our particular set of perturbations. Future studies into the mechanisms of low back pain may benefit from using a model such as ours as inputs into more sophisticated musculo-skeletal models, or may be better served by implementing equations that attempt to more realistically relate inverse dynamic measurements to low back compression (e.g. McGill, Norman, & Cholewicki, 1996).

Although this was not the main focus of our study, results from our sagittal plane lower extremity joint

angle and moment data were similar in magnitude (Table III) for individuals who ran at our three stride length conditions at $3.8 \text{ m} \cdot \text{s}^{-1}$ (Derrick *et al.*, 1998). These data have been reported here primarily to verify that the lower extremity data that contributed to our L5-S1 and T12-L1 findings were comparable with the literature.

In summary, this Newton-Euler model of the lower extremity and lumbar spine was able to detect differences in lumbo-sacral kinetics as a result of increasing stride length while running at a constant speed of $3.8 \text{ m} \cdot \text{s}^{-1}$. Specifically, L5-S1 and T12-L1 reaction forces at touchdown and during the impact phase of running increased significantly as stride length increased, as did peak L5-S1 sagittal moments and T12-L1 axial rotation moments. The L5-S1 kinematic data agree with previous literature and were sensitive enough to realise significant differences to our perturbation. Based on the results of this study, we feel that our model will be potentially useful in conducting future mechanistic research regarding L5-S1 mechanics.

Acknowledgements

This study was supported by an International Society of Biomechanics matching dissertation grant. The authors also thank Michael Moses for his assistance with data collection and processing.

References

- Andersson, G. B. (1997). The epidemiology of spinal disorders. In J. W. Frymoyer (Ed.), *The adult spine: principles and practice* (pp. 93–141). Philadelphia, PA: Lippincott-Raven.
- Armstrong, J. R. (1965). *Lumbar disc lesions*. Baltimore: Williams and Wilkins.
- Baliunas, A. J., Hurwitz D. E., Ryals, A. B., Karrar, A., Case, J. P., Block, J. A., *et al.* (2002). Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis Cartilage*, 10, 573–579.
- Callaghan, J. P., Patla, A. E., & McGill, S. M. (1999). Low back three-dimensional joint forces, kinematics, and kinetics during walking. *Clinical Biomechanics*, 14, 203–216.
- Dempster, W. T. (1955). Space requirements of the seated operator (Rep. No. Technical report WADC-TR-55-159). Ohio: Wright-Patterson Air Force Base.
- Derrick, T. R., Hamill, J., & Caldwell, G. E. (1998). Energy absorption of impacts during running at various stride lengths. *Medicine and Science in Sports and Exercise*, 30, 128–135.
- Fields, K. B., Kramer, J. S., & Delaney, M. J. (1990). Osteitis pubis and pelvic stress fracture in an elite female distance runner. *Clinical Sports Medicine*, 2, 173–178.
- Geraci, M. C., Jr. (1998). Rehabilitation of the hip, pelvis, and thigh. In W. B. Kibler, S. A. Herring, J. M. Press, & P. A. Lee (Eds.), *Functional rehabilitation of sports and musculoskeletal injuries* (pp. 216–243). Gaithersburg, MD: Aspen.
- Jacobs, S. J., & Berson, B. L. (1986). Injuries to runners: a study of entrants to a 10,000 meter race. *American Journal of Sports Medicine*, 14, 151–155.
- Khoo, B. C., Goh, J. C., & Bose, K. (1995). A biomechanical model to determine lumbosacral loads during single stance phase in normal gait. *Medical Engineering and Physics*, 17, 27–35.
- Manal, K., McClay, I., Richards, J., Galinat, B., & Stanhope, S. (2002). Knee moment profiles during walking: errors due to soft tissue movement of the shank and the influence of the reference coordinate system. *Gait and Posture*, 15, 10–17.
- McGill, S. M., Norman, R. W., & Cholewicki, J. (1996). A simple polynomial that predicts low-back compression during complex 3-D tasks. *Ergonomics*, 39, 1107–1118.
- Messier, S. P., DeVita, P., Cowan, R. E., Seay, J., Young, H. C., & Marsh, A. P. (2005). Do older adults with knee osteoarthritis place greater loads on the knee during gait? A preliminary study. *Archives of Physical Medicine and Rehabilitation*, 86, 703–709.
- Milner, C. E., Ferber, R., Pollard, C. D., Hamill, J., & Davis, I. (2006). Biomechanical factors associated with tibial stress fractures in female runners. *Medicine and Science in Sports and Exercise*, 38, 323–328.
- Nigg, B. M. (1997). Impact forces in running. *Current Opinion in Orthopedics*, 8, 43–47.
- Pearsall, D. J., Reid, J. G., & Livingston, L. A. (1996). Segmental inertial parameters of the human trunk as determined from computed tomography. *Annals of Biomedical Engineering*, 24, 198–210.
- Schache, A. G., Blanch, P., Rath, D., Wrigley, T., & Bennell, K. (2002). Three-dimensional angular kinematics of the lumbar spine and pelvis during running. *Human Movement Science*, 21, 273–293.
- Taunton, J. E., Ryan, M. B., Clement, D. B., McKenzie, D. C., Lloyd-Smith, D. R., & Zumbo, B. D. (2003). A prospective study of running injuries: the Vancouver Sun Run ‘In Training’ clinics. *British Journal of Sports Medicine*, 37, 239–244.
- Taylor, N. F., Evans, O., & Goldie, P. A. (2003). The effect of walking faster on people with acute low back pain. *European Spine Journal*, 12, 166–172.
- Vanharanta, H. (1994). The intervertebral disc: a biologically active tissue challenging therapy. *Annals of Medicine*, 26, 395–399.
- Vogt, L., & Banzer, W. (1999). Measurement of lumbar spine kinematics in incline treadmill walking. *Gait and Posture*, 9, 8–23.
- Vogt, L., Pfeifer, K., Portscher, A. M., & Banzer, W. (2001). Influences of nonspecific low back pain on three-dimensional lumbar spine kinematics in locomotion. *Spine*, 26, 1910–1919.
- Voloshin, A., & Wosk, J. (1982). An *in vivo* study of low back pain and shock absorption in the human locomotor system. *Journal of Biomechanics*, 15, 21–27.
- Whittle, M. W., Levine, D., & Pharo, E. C. (2000). Sagittal plane motion of the pelvis and lumbar spine during level, uphill and downhill walking and running. *Gait and Posture*, 11, 162.
- Winter, D. A. (1990). *Biomechanics and motor control of human movement* (2nd ed.). New York, NY: Wiley Interscience.