Differences between the sexes in the three-dimensional angular rotations of the lumbo-pelvic-hip complex during treadmill running

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Accepted 31 October 2002

The aims of this experiment were to determine whether there are differences between the sexes in the three-dimensional angular rotations of the lumbo-pelvic-hip complex during running and, if such differences exist, to establish whether factors other than sex can explain the observed differences. A cohort of 44 non-injured runners (22 males, 22 females) who usually ran more than 20 km per week were voluntarily recruited. All trials were conducted on a treadmill at a running speed of 4.0 m s⁻¹. Reflective markers were placed over specific anatomical landmarks of the thoraco-lumbar spine, pelvis and femur. Data were captured using a VICON motion analysis system. Females tended to display a greater peak-to-peak oscillation for most of the angular rotations. An offset was the main difference between the male and female group mean waveforms for pelvis anterior–posterior tilt. Forward stepwise regression analysis revealed that sex was the most common variable related to the amplitudes of the angular rotations. Given these results, sport scientists conducting future biomechanical studies using angular data to test hypotheses are advised to be extremely cautious about averaging across male and female participants.

Keywords: hip, kinematics, lumbar spine, pelvis, sex.

Introduction

The participation of females in distance running was considerably restricted during the first two-thirds of the twentieth century (Atwater, 1990). The women’s marathon only became an Olympic event in 1984 at the Games of the XXIII Olympiad in Los Angeles. With females now participating regularly in distance running, they are, like their male counterparts, presenting to sports medicine clinics with running injuries. Epidemiological studies have not found the overall incidence of running injuries to display a bias between the sexes (Jacobs and Berson, 1986; Lysholm and Wiklander, 1987; Walter et al., 1989; Bennell and Crossley, 1996). However, female runners do appear to be susceptible to different types of running injuries than male runners. For example, stress fractures of the pelvis and neck of the femur have been found to be more common in female runners (Pavlov et al., 1982; Prescott, 1983; Bennell et al., 1996). Although it is acknowledged that hormonal factors are likely to be of relevance in explaining this disparity, mechanical factors are also thought to play a role.

It has been proposed that stress fractures of the inferior pubic ramus are a result of repetitive tensile stress to the bone (Pavlov et al., 1982). The ischiopubic ramus is the origin of the obturator internus and externus as well as the adductor muscles (McMinn, 1990). The strong pull of these muscles on the lateral part of the ischiopubic ramus as the hip approaches full extension during terminal stance is believed to produce tensile stress on the medial portion of the bone (Latshaw et al., 1981; Pavlov et al., 1982). Given that stress fractures of the inferior pubic ramus were reported to occur more commonly in females than
males, Pavlov et al. (1982) suggested that female runners may rely on hip extension forces to a greater extent than male runners. They suggested that a sophisticated study of the differences in running mechanics between the sexes was required to verify this hypothesis (Pavlov et al., 1982). An understanding of the differences between the sexes in the three-dimensional angular rotations of the lumbo-pelvic-hip complex during running is, therefore, of considerable clinical interest.

Differences between the sexes in the angular rotations of the lumbo-pelvic-hip complex have been shown to occur during walking (Crosbie et al., 1997; Kerrigan et al., 1998; Stokes et al., 1989). Although it is likely that differences between the sexes would also exist during running, this has yet to be clearly established. Williams et al. (1987) compared the thigh angle of 14 elite females running at 5.4 m·s⁻¹ to that of eight males running at 5.2 m·s⁻¹. The females were found to display significantly greater maximum thigh flexion but were not found to display significantly greater maximum thigh extension. To our knowledge, differences between the sexes in the angular rotations of the lumbar spine and pelvis during running have not previously been investigated.

The aims of this experiment were two-fold. The first was to assess whether differences exist between the sexes in the three-dimensional angular rotations of the lumbo-pelvic-hip complex during running. If this was the case, the second aim was to then establish whether factors other than sex can explain the observed differences.

Materials and methods

Approval for this experiment was obtained from The University of Melbourne and The Australian Institute of Sport ethics committees. A cohort of 44 able-bodied participants (22 males, 22 females) were voluntarily recruited and written consent was obtained. Details justifying the sample size are provided in the data analysis section. Individuals were recruited if: (1) they were distance runners actively involved at either a recreational or an elite standard; (2) they usually ran on average 20 km or more a week; (3) they were not suffering from any musculoskeletal injury at the time of testing. Individuals with a history of previous injuries were not excluded. All participants had also experienced running on a treadmill before being recruited. The respective physical characteristics of the male and female participants were as follows: age, 34.6 ± 7.3 and 34.7 ± 6.1 years; body mass, 72.7 ± 6.9 and 60.8 ± 6.4 kg; height, 1.77 ± 0.05 and 1.67 ± 0.09 m.

Several pelvic anthropometric parameters were obtained. The same tester performed all measures to avoid inter-tester variability. Absolute pelvic width was measured using a standard flexible tape measure with the participant in the supine position. It was measured as the distance between the most prominent aspects of the anterior superior iliac spines. This procedure has previously been shown to be reliable (Alderink et al., 2000). Relative pelvic width was calculated by expressing absolute pelvic width as a percentage of height. Standing pelvic tilt was measured using callipers with an attached inclinometer. It was taken as the angle between the horizontal and a line connecting the anterior superior iliac spine and posterior superior iliac spine. The participants assumed a neutral standing posture. The most prominent aspects of the anterior and posterior superior iliac spines were palpated and marked. The tips of the callipers were placed over the marked locations and the angle of pelvic tilt was recorded directly from the inclinometer. The angle was positive for anterior pelvic tilt. This procedure has also previously been shown to be reliable (Walker et al., 1987; Heino et al., 1990).

All running trials were conducted on a large 15 kW treadmill (SportsTech Gymnasium and Electronic Sports Equipment, Australia), which has previously been shown to display a small variation in intra-stride belt speed (Schache et al., 2001). To familiarize the participants with the apparatus, three 10 min level treadmill runs with 5 min rest periods between runs were completed by each participant on the day preceding data collection. Minimal deviations have been found to exist in the three-dimensional angular rotations of the lumbo-pelvic-hip complex with respect to overground running when conditioned runners are well accommodated to a treadmill with a small variation in intra-stride belt speed (Schache et al., 2001).

Three-dimensional angular data were collected using a VICON 370 motion analysis system (Oxford Metrics Ltd, Oxford, UK) with seven cameras (NAC Inc., Japan) operating at a sampling rate of 200 Hz. The defined laboratory (global) orthogonal coordinate system (frame) followed the right hand rule and had the positive x-direction orientated in the direction of forward progression, the positive y-direction orientated to the left and the positive z-direction orientated vertically upwards. Errors associated specifically with the reconstruction of marker coordinates in the global frame (instrumental errors) were described by estimating the precision of the motion analysis system and were found to be no greater than 1.1°, 1.0° and 1.0° about the x, y and z axes, respectively (Schache et al., 2001, 2002a).

Reflective markers were positioned over specific anatomical landmarks of the thoraco-lumbar spine, pelvis and femur using double-sided adhesive tape (Fig. 1). The same operator performed all marker placements.
to avoid inter-tester variability. The justification for marker location, the full details regarding the methods of application and the evaluated test-retest experimental errors associated with any single measurement (intra-participant reliability) for each of the angular rotations of the lumbo-pelvic-hip complex are reported elsewhere (Schache et al., 2002a).

To define a frame at the functional upper margin of the lumbar spine, a rigid cluster consisting of three reflective markers was constructed based upon the design of Stokes (1984) and Pearcey et al. (1987). It was made from lightweight materials in an effort to minimize potential inertial artefacts. The middle of the base plate of the cluster was positioned over the twelfth thoracic spinous process (T12) and held in place by a tight elastic thoracic strap to optimize its rigidity with respect to the underlying bone.

The approximate location of the T12 spinous process was identified through palpation using a reliable technique devised by Tully and Stillman (1997) based upon a review of the surface and radiological anatomy of the lumbo-sacral region. A mark was placed on the skin at the midpoint of the posterior superior iliac spine, which overlies the second sacral spinous process (S2) (Burton et al., 1990). A second mark was placed over the spine at the level of the highest point of the iliac crest, which has been shown to correspond to the fourth lumbar spinous process (L4) (Macgibbon and Farfan, 1979). A third mark was then made on the spine superior to the second (L4) mark at a vertical distance equal to that measured between the first (S2) and the second (L4) marks. Before conducting this study, ultrasound imaging of 28 individuals with a body mass index of less than 25 found that the technique located the midpoint between the first (L1) and second (L2) lumbar spinous processes in 20 (71%) of the individuals. Hence, the technique was used in the current study safe in the knowledge that, for most participants, L1 was the spinous process immediately superior to the third mark. The T12 spinous process was then located accordingly and its position was verified by palpating along the twelfth rib medially towards the spine.

A thoraco-lumbar frame was defined using the three markers on the rigid cluster (Fig. 2). The anatomical bone-embedded frame of a vertebra has previously been defined by Pearcey (1985). As the spinous process is the only palpable anatomical landmark available for the non-invasive estimation of the in vivo angular rotations of the lumbar spine, it was not possible to reconstruct precisely the anatomical bone-embedded frame of the
vertebra in this study. Instead, with the participant in a neutral standing posture, the rigid cluster was carefully positioned over the spinous process of T12 such that its x axis was perpendicular to the skin surface in the sagittal and transverse planes and its y axis was horizontal. As none of the participants under investigation suffered from any substantial scoliotic deformities of the spine, we considered that a reasonable approximation of the anatomical bone-embedded frame of the vertebra was achieved with this configuration.

The anatomical bone-embedded frame for the pelvis has previously been defined by Cappozzo et al. (1995). The anatomical landmarks used for this frame included the left and right anterior superior iliac spines and the midpoint between the two posterior superior iliac spines. As these anatomical landmarks can all be identified through palpation, reflective markers were placed over both anterior superior iliac spines and the midpoint between the two posterior superior iliac spines and were used to define a pelvic frame (Fig. 2) that was consistent with the proposed anatomical bone-embedded frame.

Finally, for the femur, a reflective marker was placed on the lateral femoral epicondyle and a reflective marker fixed to a lightweight wand was positioned over the lateral aspect of the distal third of the thigh. A femoral frame (Fig. 2) that approximated the proposed anatomical bone-embedded frame for the femur (Cappozzo et al., 1995) was defined using these markers together with estimations of internal anatomical landmarks such as the femoral head (or hip joint centre) and the midpoint between the medial and lateral femoral epicondyles (or knee joint centre), as has been previously described (Schache et al., 2001, 2002a).

The participants were required to run in their own running shoes and to wear low-rise running shorts that did not obscure the anterior or mid-posterior superior iliac spine landmarks. The females also wore a sports top that allowed exposure of the thoraco-lumbar spine. All participants were tested at a running speed of 4.0 m:s⁻¹. After a brief-warm up at the test speed, the three-dimensional trajectories of the markers were captured. Five seconds of data were collected for each participant, which approximated six complete running cycles.

Data analysis

Data reduction

Lumbar spine movement was defined as movement of the thoraco-lumbar frame with respect to the pelvic frame. It therefore represented an estimation of the overall sum of each of the inter-vertebral movements combined. No individual inter-vertebral angular infor-

mation was available from the marker configuration utilized. Pelvic movement was defined as movement of the pelvic frame with respect to the global frame. Hip movement was defined as movement of the femoral frame with respect to the pelvic frame. The neutral position (i.e. nil rotation values on all axes) corresponded to the case where the two frames used to define a joint or segment were aligned.

Three-dimensional lumbar spine and hip angular data were calculated using a joint coordinate system as described by Grood and Suntay (1983). The International Society of Biomechanics has recently recommended this geometrical convention as the standard for the three-dimensional angular description of regional motion of the spine and motion of the hip joint (Wu et al., 2002). In the present study, the medial-lateral (y) axis of the joint coordinate system was defined as the y axis of the reference or fixed segment frame. The longitudinal (z) axis of the joint coordinate system was defined as the z axis of the moving segment frame. The ‘floating’ anterior–posterior (x) axis was defined as the common perpendicular to the y and z axes at each given instant in time. This joint coordinate system is mathematically equivalent to Cardan angles obtained from three sequential rotations performed about each of the axes of the moving segment frame in the order: (1) rotation about the y axis, followed by (2) rotation about the x axis, followed by (3) rotation about the z axis (Cole et al., 1993; Woltring, 1994). In anatomical terms, for the lumbar spine this order corresponds to (1) flexion–extension, followed by (2) lateral bend, followed by (3) axial rotation; and for the hip (1) flexion–extension, followed by (2) adduction–abduction, followed by (3) axial rotation.

Three-dimensional pelvis angular data were calculated using a segment coordinate system and movement was described with respect to the global frame. Cardan angles were obtained from three sequential rotations performed about each of the axes of the moving pelvic frame in the order: (1) rotation about the y axis, followed by (2) rotation about the x axis, followed by (3) rotation about the z axis. In anatomical terms, for the pelvis this order corresponds to (1) anterior–posterior tilt followed by (2) obliquity, followed by (3) axial rotation.

The events of initial contact and toe-off were determined from the kinematic data using both the vertical displacement and vertical velocity of additional markers placed on both lateral malleoli and the distal ends of both second metatarsals immediately adjacent to the second metatarso-phalangeal joints. This allowed the absolute spatio-temporal parameters of stride time, stride rate, stride length, stance time and swing time to be measured. Angular data for each running cycle, from right initial contact to the following right initial contact,
were time normalized to 101 points representing intervals from 0 to 100%. Repeated running cycles were averaged for each participant. Several spatio-temporal parameters were also recalculated as relative measures using an ad hoc scaling strategy (Pieryn:owski and Galea, 2001). Stride length was expressed as a proportion of height. Rather than scaling stance and swing time by (height)^1/2, both were instead expressed as a proportion of the total running cycle (stride).

**Statistical analysis**

All analyses were performed with the Statistical Package for Social Sciences (SPSS Inc., Chicago, IL). A two-tailed level of significance was set at 0.05 for all tests unless otherwise specified.

**Comparison of waveforms**

To allow statistical comparison of the group mean waveforms for a given angular rotation, the differences between the group mean waveforms ± 95% confidence limits were calculated and plotted over the running cycle. When calculating any of the confidence limits, the critical value of t was obtained with the number of degrees of freedom associated with the unpaired t-test adjusted so as to not assume homogeneity of variance. Group mean waveforms for each of the angular rotations of the lumbo-pelvic-hip complex were smoothed by fitting a 15th-order polynomial using the software program Igor Pro (Wavemetrics, Lake Oswego, OR) before plotting. Compared with other polynomial equations, the 15th-order polynomial created a curve that visually best fitted the raw group mean waveforms for each of the angular rotations.

**Comparison of measured parameters**

The anthropometric parameters of interest were height, pelvic width, relative pelvic width and standing pelvic tilt. The spatio-temporal parameters of interest were stride time, stride rate, stride length, relative stride length, stance time, swing time, relative stance time and relative swing time. The angular parameters of interest were the amplitudes of each of the angular rotations of the lumbo-pelvic-hip complex. The amplitude represented the entire range of excursion for a given angular rotation. Preliminary detailed examination of the data, including values for skewness and kurtosis together with Shapiro-Wilk tests, revealed that the statistical assumptions of normality were met, allowing parametric tests to be used.

The parameters for males and females were compared using an unpaired t-test. Pearson’s r values were calculated to determine which anthropometric and spatio-temporal parameters were significantly correlated with the amplitudes of each of the angular rotations of the lumbo-pelvic-hip complex. Forward stepwise regression analysis was performed to determine which parameters were predictive of the amplitudes of the angular rotations. The dependent variable was the amplitude of a given angular rotation of the lumbo-pelvic-hip complex and the independent variables included sex, together with other anthropometric and spatio-temporal parameters found to display either a significant difference between the sexes or a significant univariate correlation with the amplitude of a given angular rotation. As it has been recommended that at least five times more participants than independent variables are required when using regression analysis (Tabachnick and Fidell, 1989), the total number of parameters selected as potential independent variables did not exceed a maximum of eight at any time.

**Power analysis**

Power calculations were performed before the start of the study to estimate the necessary sample size. The means and standard deviations for the amplitudes of each of the angular rotations of the lumbo-pelvic-hip complex for 20 males running at a speed of 4.0 m s⁻¹ were obtained from the sample reported in Schache et al. (2002b). The anticipated minimum effect size for each of the angular rotations of the lumbo-pelvic-hip complex was represented by the magnitude of the error (maximum root mean square) associated with measuring the respective angular rotations during running (see Schache et al., 2002a). Using these data, the sample size of 44 provided at least 80% power of finding a significant difference between the sexes for angular rotations of the lumbo-pelvic-hip complex that was considered to be meaningful.

**Results**

Significant differences between males and females were found for some of the anthropometric parameters (Table 1). Males were significantly taller and had a smaller standing pelvic tilt angle than females. No significant differences between males and females were apparent for absolute or relative measures of pelvic width.

When running at a speed of 4.0 m s⁻¹, significant differences were found between males and females for all of the absolute spatio-temporal parameters (Table 2). Females displayed a shorter stance time, swing time, stride time and stride length, and a higher stride rate than males. However, none of the relative spatio-
Table 1. Anthropometric differences between male and female runners (mean ± s)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Males</th>
<th>Females</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Height (m)</td>
<td>1.77±0.05</td>
<td>1.67±0.09</td>
<td>0.0001*</td>
</tr>
<tr>
<td>Absolute pelvic width (cm)</td>
<td>24.3±1.5</td>
<td>23.5±1.9</td>
<td>0.124</td>
</tr>
<tr>
<td>Relative pelvic width (% height)</td>
<td>13.7±0.9</td>
<td>14.1±1.2</td>
<td>0.229</td>
</tr>
<tr>
<td>Standing pelvic tilt (°)</td>
<td>5.4±3.3</td>
<td>8.7±3.8</td>
<td>0.004*</td>
</tr>
</tbody>
</table>

*Significant difference (P < 0.05).

Table 2. Absolute and relative measures of spatio-temporal parameters for males and females running at 4.0 m·s⁻¹ (mean ± s)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Males</th>
<th>Females</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride time (s)</td>
<td>0.70±0.04</td>
<td>0.66±0.03</td>
<td>0.001*</td>
</tr>
<tr>
<td>Stride rate (strides·s⁻¹)</td>
<td>1.43±0.09</td>
<td>1.52±0.07</td>
<td>0.001*</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>2.80±0.17</td>
<td>2.63±0.13</td>
<td>0.001*</td>
</tr>
<tr>
<td>Relative stride length (% height)</td>
<td>157.9±9.7</td>
<td>158.0±8.3</td>
<td>0.952</td>
</tr>
<tr>
<td>Stance time (s)</td>
<td>0.23±0.01</td>
<td>0.22±0.01</td>
<td>0.012*</td>
</tr>
<tr>
<td>Swing time (s)</td>
<td>0.47±0.04</td>
<td>0.44±0.03</td>
<td>0.004*</td>
</tr>
<tr>
<td>Relative stance time (% stride)</td>
<td>32.8±2.2</td>
<td>33.2±1.5</td>
<td>0.438</td>
</tr>
<tr>
<td>Relative swing time (% stride)</td>
<td>67.2±2.2</td>
<td>66.8±1.5</td>
<td>0.438</td>
</tr>
</tbody>
</table>

* Significant difference (P < 0.05).

temporal parameters were found to be significantly different. Relative stride length, relative stance time and relative swing time were found to be similar for males and females.

The male and female group mean waveforms for the angular rotations of the lumbo-pelvic-hip complex during running are displayed in Figs 3–5. The magnitude and timing of differences between the male and female group mean waveforms over the running cycle are illustrated in Figs 6–8. Differences between the male and female group mean waveforms were observed in the peak-to-peak oscillations, with females tending to display greater amplitudes. The patterns of the male and female group mean waveforms for a given angular rotation were remarkably similar. The waveforms for lumbar spine lateral bend and axial rotation, pelvis obliquity and axial rotation, and hip adduction–abduction demonstrate this finding best. For these angular rotations, the greatest deviations from zero tended to occur at approximately the same time in the running cycle as the peak reversal points on the corresponding waveform. For example, the peak reversal points for pelvis axial rotation occurred at approximately 15 and 65% of the running cycle (Fig. 4c). The difference between the group means ± 95% confidence limits plot deviated from zero at these same times in the running cycle (Fig. 7c). At 15% of the running cycle, the bias indicates that the waveform for males was greater than the waveform for females. Upon visual inspection of the pattern for pelvis axial rotation, the waveforms were both negative at 15% of the running cycle (Fig. 4c). The waveform for males was, therefore, less negative than the waveform for females, meaning that the absolute magnitude of the peak reversal point in the waveform for males was less than that for females. The same finding occurred at 65% of the running cycle, except that the waveforms for males and females were both positive at this time.

For pelvis anterior–posterior tilt, an offset appeared to be the main difference between the male and female group mean waveforms (Fig. 4a). The mean position of anterior pelvic tilt across the running cycle averaged 20.2±4.0° (mean ± s) for females and 16.9±4.3° for males. This offset for pelvis anterior–posterior tilt was found to be significantly different between the sexes (P=0.013) and is clearly displayed by the plot of the difference between the group means ± 95% confidence limits, where a consistent negative bias (females > males) across the running cycle is evident (Fig. 7a).

Differences between the male and female group mean waveforms for hip flexion–extension were also observed (Fig. 5a). Females tended to commence hip flexion during initial swing at an earlier stage than males. Females also tended to have a greater angle of peak hip flexion during the later third of swing than males. These findings are well illustrated by the plot of the difference between the group means (Fig. 8a), where both the upper and lower 95% confidence limits surrounding the difference between the group means were found to be less than zero, indicating a negative bias (females > males) during initial swing and the later third of swing.

Given that one of the main differences evident between the group mean waveforms for males and females was the peak-to-peak oscillation, the amplitudes of each of the angular rotations of the lumbo-pelvic-hip complex for males and females were calculated and directly compared. Females displayed significantly greater amplitudes of lumbar spine lateral bend and axial rotation, pelvis anterior–posterior tilt, obliquity and axial rotation, and hip adduction–abduction than their male counterparts (Table 3). The magnitude of the differences (± 95% confidence intervals) ranged from 1.6° (s = 1.3°) for pelvis anterior–posterior tilt to 6.6° (s = 3.0°) for hip adduction–abduction. No significant differences between the sexes were found for...
the amplitudes of lumbar spine flexion-extension, hip flexion-extension and hip axial rotation.

Forward stepwise regression analysis was used to determine which factors independently accounted for a significant proportion of the variance in the amplitudes of each of the angular rotations of the lumbo-pelvic-hip complex. Using the criteria previously outlined, the measured parameters selected as independent variables included sex, height, standing pelvic tilt, stride rate, stride length, stance time and swing time. Other parameters not found to be significantly different between the sexes but found to display a significant univariate correlation with a given angular rotation were also included. Relative stride length was included as an additional independent variable for predicting the amplitude of lumbar spine flexion-extension and pelvic anterior–posterior tilt, while relative pelvic width was included as an additional independent variable for predicting the amplitude of lumbar spine axial rotation. Thus, for these three angular rotations, eight parameters were included as independent variables in the regression analysis.

None of the chosen independent variables were found to be predictive of the variance in the amplitudes of hip flexion–extension and hip axial rotation (Table 4). For the other angular rotations, sex was found to be the variable that was most able to explain a significant independent proportion of the variance in the amplitudes.
Fig. 5. Group mean waveforms for the angular rotations of the hip joint over the running cycle from right initial contact (RIC) to right initial contact. The solid line represents females; the dashed line represents males. Toe-off is indicated by the thin vertical dashed line.

Fig. 6. Differences between the sexes in the three-dimensional angular rotations of the lumbar spine at a running speed of 4.0 m s⁻¹. The differences between the independent group means (solid line) ± 95% confidence limits (dashed lines) are plotted over the running cycle for each of the lumbar spine angular rotations. Toe-off is indicated by the thin vertical dashed line. M = males; F = females; RIC = right initial contact.

Discussion

Although the overall incidence of running injuries has not been found to display any bias between the sexes (Jacobs and Berson, 1986; Lysholm and Wiklander, 1987; Walter et al., 1989; Bennell and Crossley, 1996), certain types of injuries, such as pelvic-femoral stress fractures, have been reported to be more prevalent in female runners (Pavlov et al., 1982; Prescott, 1983; Bennell et al., 1996). Potential differences in running mechanics between the sexes could be of relevance in this context (Pavlov et al., 1982). The aim of the present study, therefore, was to determine whether there are differences between the sexes in the three-dimensional angular rotations of the lumbo-pelvic-hip complex during running.

Standing pelvic tilt angle was the only pelvic anthropometric parameter measured that displayed a significant difference between the sexes (Table 1). Females were found to have a greater standing pelvic tilt angle than males. This concurs with previous findings in the literature (Youdas et al., 1996). Both absolute and relative measures of pelvic width were similar between males and females in the present study, which is consistent with the results of Williams et al. (1987). These findings suggest that female runners do
not have wider pelves than males. This contrasts with what has been suggested in the clinical literature (Potera, 1986).

Relative stride length did not differ between males and females (Table 2). As females were shorter than males, this meant that the absolute stride length of females was less than that of males. Females, therefore, adopted a faster stride rate to run at the same absolute speed as their male counterparts. Some previous investigations have reported similar findings for step length (Nelson et al., 1977; Elliott and Blanksby, 1979; Lui, 1984), stride length (Bhambhani and Singh, 1985) and step rate (Nelson et al., 1977; Elliott and Blanksby, 1979; Lui, 1984). Others, however, have reported conflicting results for step length and step rate (Elliott and Blanksby, 1976), relative step length (Nelson et al., 1977) and relative stride length (Williams et al., 1987) that may be due to differences in participant demographics and experimental running speeds.

The females in the present study were found to have shorter stance and swing times than males (Table 2). However, the stance and swing phases consisted of the same proportion of the running cycle. For both groups, toe-off occurred at approximately one-third of the
running cycle. In line with these findings, all previous investigations have found females to display a shorter stance time (Elliott and Blanksby, 1976; Nelson et al., 1977; Lui, 1984). However, mixed results have been reported for swing time (Elliott and Blanksby, 1976; Lui, 1984).

For certain angular rotations, such as lumbar spine lateral bend, lumbar spine axial rotation and pelvis anterior–posterior tilt, the magnitude of the significant difference between the group means in the current study approximated the respective magnitude of the previously evaluated test–retest experimental error for a single measurement (see Schache et al., 2002a). This means that it cannot be guaranteed that the differences found between males and females will be upheld for the single measurement of any given individual. It does not mean, however, that the differences between the males and females do not represent real differences.

Most of the observed differences between the sexes in the angular rotations related to either variations in amplitude or the presence of an offset between the group mean waveforms. Hip axial rotation was the only angular rotation that did not appear to display a difference between the sexes (Figs 5c and 8c). This may be because there is no true difference between the sexes in hip axial rotation during running. Alternatively, a true difference between males and females for hip axial rotation may exist, but the possible errors associated with measuring hip axial rotation with the current marker set-up (Schache et al., 2002a) may have prevented the true difference from being detected.

For lumbar spine lateral bend (Fig. 3b) and axial rotation (Fig. 3c), pelvis obliquity (Fig. 4b) and axial rotation (Fig. 4c), and hip adduction–abduction (Fig. 5b), the most prominent difference pertained to the peak-to-peak oscillation, with females displaying a greater amplitude than males. This finding is supported by several factors. First, visual inspection of the group mean waveforms for these angular rotations revealed very similar basic patterns between males and females. Secondly, the plots of the differences between the group means ±95% confidence limits for lumbar spine lateral bend (Fig. 5b) and axial rotation (Fig. 6c), pelvis obliquity (Fig. 7b) and axial rotation (Fig. 7c), and hip adduction–abduction (Fig. 8b) demonstrate that most of the deviations from zero tended to occur close to the times of the absolute peak magnitudes in the respective angular waveforms. Finally, a comparison of the amplitudes of these angular rotations between males and females yielded statistically significant differences (Table 3).

Even though the amplitude of pelvis anterior–posterior tilt was also found to be significantly different between males and females (Table 3), the most prominent difference between the sexes for this angular rotation related to the presence of an offset between the group mean waveforms (Fig. 4a). This is confirmed by the consistent negative bias (females > males) present in the plot of the difference between the group means ±95% confidence limits (Fig. 7a). Females were found to run with a significantly increased mean position of anterior pelvic tilt over the running cycle compared with males. Although it is acknowledged that the measurement of anterior tilt of the pelvis is highly susceptible to errors associated with the placement of the reflective marker over the mid-posterior superior iliac crest.

### Table 3. Average amplitudes (in degrees) of each of the angular kinematic parameters of the lumbo-pelvic-hip complex for males and females running at 4.0 m·s⁻¹

<table>
<thead>
<tr>
<th>Kinematic parameter</th>
<th>Males</th>
<th>Females</th>
<th>P</th>
<th>Difference</th>
<th>95% CI</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Lumbar spine</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(y) Flexion–extension</td>
<td>12.8</td>
<td>15.3</td>
<td>0.073</td>
<td>−2.4</td>
<td>±2.6</td>
</tr>
<tr>
<td>(x) Lateral bend</td>
<td>19.3</td>
<td>22.2</td>
<td>0.027*</td>
<td>−3.0</td>
<td>±2.6</td>
</tr>
<tr>
<td>(z) Axial rotation</td>
<td>23.6</td>
<td>28.5</td>
<td>0.006*</td>
<td>−4.9</td>
<td>±3.4</td>
</tr>
<tr>
<td><strong>Pelvis</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(y) Anterior–posterior tilt</td>
<td>7.8</td>
<td>9.4</td>
<td>0.015*</td>
<td>−1.6</td>
<td>±1.3</td>
</tr>
<tr>
<td>(x) Obliquity</td>
<td>13.8</td>
<td>19.3</td>
<td>0.0001*</td>
<td>−5.5</td>
<td>±2.2</td>
</tr>
<tr>
<td>(z) Axial rotation</td>
<td>13.3</td>
<td>18.3</td>
<td>0.002*</td>
<td>−5.0</td>
<td>±3.1</td>
</tr>
<tr>
<td><strong>Hip</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(y) Flexion–extension</td>
<td>70.6</td>
<td>74.3</td>
<td>0.079</td>
<td>−3.7</td>
<td>±4.1</td>
</tr>
<tr>
<td>(x) Adduction–abduction</td>
<td>23.1</td>
<td>29.4</td>
<td>0.0001*</td>
<td>−6.6</td>
<td>±3.0</td>
</tr>
<tr>
<td>(z) Axial rotation</td>
<td>36.6</td>
<td>32.6</td>
<td>0.134</td>
<td>3.7</td>
<td>±4.8</td>
</tr>
</tbody>
</table>

*Negative value implies males < females; CI = confidence interval. *Significant difference (P < 0.05).
Table 4. Results of the forward stepwise regression analyses

<table>
<thead>
<tr>
<th>Dependent variable</th>
<th>Independent variable</th>
<th>$\beta$</th>
<th>Standard error</th>
<th>$P$</th>
<th>Variance explained (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lumbar spine</td>
<td>Constant</td>
<td>-15.486</td>
<td>11.454</td>
<td>0.184</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>Relative stride length</td>
<td>0.187</td>
<td>0.072</td>
<td>0.013</td>
<td>13.7</td>
</tr>
<tr>
<td>(x) Lateral bend</td>
<td>Constant</td>
<td>16.302</td>
<td>2.036</td>
<td>0.000</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>Sex</td>
<td>2.056</td>
<td>1.287</td>
<td>0.027</td>
<td>11.1</td>
</tr>
<tr>
<td>(z) Axial rotation</td>
<td>Constant</td>
<td>-23.328</td>
<td>15.728</td>
<td>0.146</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>Sex</td>
<td>4.858</td>
<td>1.792</td>
<td>0.010</td>
<td>16.6</td>
</tr>
<tr>
<td></td>
<td>Stance time</td>
<td>171.218</td>
<td>64.596</td>
<td>0.011</td>
<td>7.9</td>
</tr>
<tr>
<td></td>
<td>Standing pelvic tilt</td>
<td>0.540</td>
<td>0.231</td>
<td>0.025</td>
<td>9.1</td>
</tr>
<tr>
<td>Pelvis</td>
<td>Constant</td>
<td>81.162</td>
<td>22.881</td>
<td>0.001</td>
<td>–</td>
</tr>
<tr>
<td>(y) Anterior–posterior tilt</td>
<td>Sex</td>
<td>2.947</td>
<td>0.568</td>
<td>0.000</td>
<td>13.3</td>
</tr>
<tr>
<td></td>
<td>Stride rate</td>
<td>-35.148</td>
<td>9.074</td>
<td>0.000</td>
<td>26.0</td>
</tr>
<tr>
<td></td>
<td>Swing time</td>
<td>-54.933</td>
<td>21.902</td>
<td>0.016</td>
<td>8.2</td>
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<tr>
<td>(x) Obliquity</td>
<td>Constant</td>
<td>8.322</td>
<td>1.703</td>
<td>0.000</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>Sex</td>
<td>5.464</td>
<td>1.077</td>
<td>0.000</td>
<td>38.0</td>
</tr>
<tr>
<td>(z) Axial rotation</td>
<td>Constant</td>
<td>9.160</td>
<td>2.407</td>
<td>0.000</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>Sex</td>
<td>6.524</td>
<td>1.655</td>
<td>0.000</td>
<td>20.2</td>
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<tr>
<td></td>
<td>Standing pelvic tilt</td>
<td>-0.446</td>
<td>0.216</td>
<td>0.045</td>
<td>7.5</td>
</tr>
<tr>
<td>Hip</td>
<td>Nil entered</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>(y) Flexion–extension</td>
<td>Constant</td>
<td>-17.778</td>
<td>13.719</td>
<td>0.202</td>
<td>–</td>
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<tr>
<td>(x) Adduction–abduction</td>
<td>Sex</td>
<td>7.762</td>
<td>1.514</td>
<td>0.000</td>
<td>29.9</td>
</tr>
<tr>
<td></td>
<td>Stance time</td>
<td>144.894</td>
<td>56.724</td>
<td>0.014</td>
<td>9.6</td>
</tr>
</tbody>
</table>

Note: $\beta$ = unstandardized coefficients.

(Schache et al., 2002a), it is unlikely that such errors influenced this finding. Errors associated with the placement of the reflective marker over the mid-posterior superior iliac crest would have been equal for male and female pelvises and random in nature. Such errors cannot explain the systematic increase in anterior pelvic tilt during running for females. This result may be a product of the greater standing pelvic tilt angle of female runners in this study. Several clinical reports have suggested that increased anterior pelvic tilt during running is related to injury (Klein and Roberts, 1976; Geraci, 1996; Bruker and Khan, 2001). Whether increased anterior pelvic tilt is, therefore, a relevant mechanical factor in explaining the apparent susceptibility of female runners to pelvic-femoral stress fractures requires further verification.

Differences between the male and female group mean waveforms for hip flexion–extension were observed during the initial and later stages of the swing phase during the running cycle (Figs 5a and 8a). Females tended to commence hip flexion immediately before toe-off, whereas males remained in hip extension for a short period during the initial swing. Females also tended to have a greater angle of peak hip flexion during the later third of swing than males. This is in line with the results of Williams et al. (1987), who found females to display significantly greater maximum thigh flexion than males.

The prevalence of pelvic-femoral stress fractures in female military recruits has been suggested to be a direct result of repetitive 'overstriding' when attempting to march in time with their taller male counterparts (Ozburn and Nichols, 1981; Hill et al., 1996). This biomechanical hypothesis is also thought to be a possible explanation for the prevalence of pelvic-femoral stress fractures in female distance runners. Pavlov et al. (1982) proposed that female runners rely on hip extension forces to a greater extent than male runners. However, when running at the same absolute speed, females in the current study did not appear to 'overstride' compared with the males. Relative stride length was not found to be significantly different between males and females (Table 2). In addition, the hip flexion–extension waveform for females did not
display a significantly increased amplitude with respect to males (Fig. 5a) as would be expected if females were ‘overstriding’). Consequently, these results do not provide any evidence to support the theory proposed by Pavlov et al. (1982). To further test this theory, future studies should examine differences between the sexes in joint moments and powers during running.

The reason for the observed differences in the angular rotations between males and females in this study is likely to be multi-factorial. In other words, sex may not be the only factor capable of explaining a significant proportion of the measured variance in the amplitudes of the angular rotations of the lumbo-pelvic-hip complex during running. Other factors, such as the postural configuration of the pelvis and running stride characteristics, could be relevant. Although running speed was fixed, each participant was free to self-select his or her own stride length and stride rate. Significant differences were found between males and females for all of the absolute spatio-temporal parameters (Table 2). Williams et al. (1987) reported that spatio-temporal parameters influence angular movements during running. For example, a longer stride length was related to increased hip extension ($r = 0.63$). Thus, differences between the sexes in the amplitudes of the angular rotations evident in this study could merely be a product of the significant differences in self-selected spatio-temporal parameters.

To address the above possibilities, forward stepwise regression analysis was used, in which sex and several selected anthropometric and spatio-temporal variables were included as independent variables. Height failed to obtain statistical significance and so was not entered as an independent variable in the regression equation for any of the angular rotations (Table 4). Therefore, height was not considered to be much of an influence on the amplitudes of the angular rotations of the lumbo-pelvic-hip complex during running. This result is consistent with previously described scaling strategies, which suggest that angles are effectively dimensionless (Hof, 1996) and remain independent of body size parameters (Pieyruski and Galea, 2001). With respect to the pelvic anthropometric parameters included as independent variables, standing pelvic tilt was the only one found to be a significant predictor. It was able to explain a significant proportion of the variance in the amplitudes of both lumbar spine and pelvis axial rotation independent of sex. Thus, axial rotation within the lumbo-pelvic region does appear to be related to the postural configuration of the pelvis. When determining whether any of the measured spatio-temporal parameters were predictive of the variance of the amplitudes of the angular rotations of the lumbo-pelvic-hip complex during running, mixed results were obtained and were difficult to interpret. Relative stride length was predictive of the amplitude of lumbar spine flexion-extension, stride rate and swing time were predictive of the amplitude of pelvis anterior-posterior tilt, while stance time was predictive of the amplitude of lumbar spine axial rotation and hip adduction-abduction (Table 4). Consequently, it would appear that the amplitudes of some of the angular rotations of the lumbo-pelvic-hip complex during running are influenced by certain spatio-temporal parameters.

Even though standing pelvic tilt and several spatio-temporal parameters were found to be predictor variables for the amplitudes of certain angular rotations, forward stepwise regression analysis revealed that sex was the most common variable related to the amplitudes of the angular rotations of the lumbo-pelvic-hip complex during running. Sex was able to explain a significant proportion of the amplitudes of lumbar spine lateral bend and axial rotation, pelvis anterior-posterior tilt, obliquity and axial rotation, and hip adduction-abduction (Table 4). An inherent difference between the sexes, therefore, appears to exist in the amplitudes of most of the angular rotations of the lumbo-pelvic-hip complex during running.

One factor not quantified in this experiment but potentially of relevance is running speed expressed relative to maximum speed. The maximum running speed of males has previously been shown to be significantly greater than that of females (Nelson et al., 1977). If it is assumed that this was also true for the males and females in the present study, then the test speed of 4.0 m·s$^{-1}$ represented a faster running speed relative to maximum for females. Nelson et al. (1977) compared numerous biomechanical parameters for males and females running at the same speed relative to maximum. Although the differences between males and females were not eliminated, they were found to change from those present when males and females were compared at the same absolute running speed.

To assess whether relative running speed influenced the differences in the amplitudes of the angular rotations observed in the present study, it would have been necessary to have expressed the test speed of 4.0 m·s$^{-1}$ as a proportion of maximum speed for each participant, and then used this as an independent variable in the regression analysis. However, this required each participant’s maximum running speed to be quantified. Given that the cohort in this investigation comprised runners of varying ability undertaking differing amounts of training at the time of testing, we considered it too difficult to experimentally measure maximum running speed in a truly valid manner.

It is important to note that the observations made in this experiment were limited to treadmill running. Treadmill running was chosen due to the well-standardized and reproducible running environment provided.
This was considered important to allow the males and females to be compared under identical running conditions. Furthermore, when comparing treadmill and overground running, few differences have been found to exist in the three-dimensional angular rotations of the lumbo-pelvic-hip complex (Schache et al., 2001).

The results of the present study have an important implication for future studies investigating running biomechanics. Although the basic patterns of the angular rotations of the lumbo-pelvic-hip complex during running were found to be quite similar between males and females, significant differences were noted for the amplitudes of some of the angular rotations. Other studies have found significant differences to exist between the sexes in knee and ankle angular data during running (Williams et al., 1987; Dutto et al., 1997). Consequently, sport scientists conducting future biomechanical studies using angular data to test hypotheses are advised to be extremely cautious about averaging across male and female participants.

Acknowledgements

We are greatly appreciative of the support for this project provided by L. Philpot of the Australian Institute of Sport and D. Hopkins of Nike Australia.

References


