Review article

The coordinated movement of the lumbo–pelvic–hip complex
during running: a literature review

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Abstract

The purpose of this review article is to summarise the literature to date regarding the movement of the lumbar spine, pelvis and hips during running gait. Both two-dimensional and three-dimensional studies are analysed to illustrate the apparent coordination in the angular kinematics of each of these segments during running. Knowledge of this coordination is essential in order to facilitate the successful rehabilitation of running injuries to the back, pelvis, hip and thigh. © 1999 Elsevier Science B.V. All rights reserved.

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1. Introduction

The potential benefits of regular physical exercise are well known. Being relatively convenient and inexpensive, running is a popular sport at both recreational and competitive levels. Unfortunately, this has been associated with an increase in the number of running injuries [1]. Injuries pertaining to the back, pelvis, hip and thigh account for approximately 25–35% of all injuries sustained by distance runners and sprinters of varying levels of ability [2–5]. It is clearly evident from the epidemiological literature that injuries to this region in the running population are not as common as those around the knee and lower leg complex. However, some of the overuse injuries that occur around the back and the pelvis are frequently debilitating requiring prolonged periods of rehabilitation [6–8]. It is also not unusual for injuries of the thigh, such as hamstring strains in sprinters, to become recurrent [9–11]. Thus, injuries to the back, pelvis, hip and thigh in the running population can be a potential source of frustration for the clinician. This demands the identification of risk factors for these injuries in order to facilitate the implementation of effective preventative measures.

Biomechanical evaluation of running has the potential to identify risk factors and contribute to the prevention of running injuries [12,13]. Most of the biomechanical research aiming to reduce the frequency of running injuries has focused on the lower leg complex. Many studies have assessed the coordinated movement of the knee, ankle and subtalar joints during running [14–23]. The possible relationship between alterations to the coordinated movement of the lower leg complex during running and injury has also been investigated [19,20,22,24–29]. In comparison, research on the integrated biomechanical function of the lumbo–pelvic–hip complex during running is relatively scarce. Given the intimate anatomical connections that exist between the spine, pelvis and femur [30–33], it is likely that this complex will also be highly coordinated during running. Evidence is available suggesting this is the case for walking [34–43]. Knowledge of this coordination during running is essential in order to facilitate the
successful rehabilitation of running injuries to the back, pelvis, hip and thigh.

The first authors to emphasise the biomechanical function of the lumbar spine and pelvis during running were Slocum and Bowerman [44] who described the position of the pelvis to be the key to postural control. Since then interest in the biomechanical function of this region during running has continued to develop, especially through the work of Gracovetsky and colleagues [41,45–49]. Gracovetsky’s hypotheses, as will be discussed, certainly stimulate sufficient controversy to warrant further research into the biomechanical interaction between of the lumbar spine, pelvis and hip during running.

2. Rotational kinematics of the lumbo–pelvic–hip complex during running

The aim of this review is to examine the kinematic coordination within the lumbo–pelvic–hip complex during running. In doing so, several points need to be noted. Firstly, whilst absolute linear displacement of each of the body segments can be expected to occur, it has only been measured in limited circumstances during running [50]. Therefore, this review focuses only on rotational kinematics during running, as the degree of information available for integration is considerably greater. Secondly, to date, published data detailing the rotational kinematics of the lumbar spine during running are not available. Therefore, data on the rotational kinematics of the trunk are used to provide an indication of lumbar spine movement patterns during running.

The movement within the lumbo–pelvic–hip complex will be described with respect to the phases of the running cycle. One complete running cycle for a given lower extremity is made up of stance and swing phases. The stance phase can be subdivided into the periods of absorption and propulsion. Absorption commences at foot strike and finishes at midstance, whilst propulsion commences at midstance and finishes at toe off. The swing phase can also be subdivided into the periods of initial and terminal swing. Initial swing commences at toe off and finishes at midswing, whilst terminal swing commences at midswing and finishes at foot strike. The stance and swing phases comprise approximately 40 and 60% of the running cycle respectively. This means that an airborne period where both feet are off the ground (approximately 10% of the running cycle) occurs at the beginning and the end of each swing phase [51].

For the purpose of describing the rotational kinematics of the lumbo–pelvic–hip complex, this area can be subdivided into three body segments: the trunk, the pelvis and the thigh. Detailing the rotational kinematics then involves measuring either segment angles or joint angles. Segment angles are calculated by measuring the motion of a single body segment with respect to an external reference (measurement in a global coordinate system). Joint angles are calculated by measuring the relative motion between two adjacent body segments.

Segment or joint angles can be measured using either two dimensional (2D) or three dimensional (3D) analysis techniques. 2D analysis involves filming one side of the body only and represents the projection of data onto a single viewing plane. 3D analysis involves modeling the body segments as rigid bodies. Kinematic data are calculated by comparing the rotation about three orthogonal axes of one particular body segment with respect to a reference set of axes. The reference set of axes correspond to the laboratory (global) axes when measuring segment angles, or the adjacent proximal body segment axes when measuring joint angles. For further information on the modeling procedures that have commonly been used in 3D analysis, the reader is referred to the work of Davis et al. [52,53], Cappozzo [54,55] and Kadaba et al. [56].

Most authors have defined the trunk as a single segment [50,57–69]. Segment angles have then been used to describe the rotational kinematics of the trunk during running with respect to either the vertical [57,58,60–64,66–69] or the angle of the trunk in a neutral standing posture [50,59]. As well as defining the trunk as a single segment, Thorstensson and colleagues [50] also divided the trunk into two segments in the coronal plane and measured a mid-trunk joint angle. Most studies have used 2D analysis to measure the rotational kinematics of the trunk during running [50,57–65,67,69], with few using 3D analysis [66,68].

Segment angles have also typically been used to describe the rotational kinematics of the pelvis. For running, this has become more common with the recent proliferation of commercially available 3D measurement systems, as simple 2D analysis is likely to yield misleading information. Gard et al. [70] have illustrated this for walking using 2D Lissajous plots of the coronal plane movement of markers on the anterior and posterior aspects of the pelvis. They showed that transverse plane rotation of the pelvis during walking results in apparent medial and lateral translation of the anterior and posterior markers on the pelvis when the data is projected into the coronal plane. 2D analysis of the coronal plane translations of the markers on the pelvis during walking was demonstrated to be a combined measure of actual pelvic translations in the medial–lateral and vertical directions as well as apparent translations due to the rotation of the pelvis in the transverse plane.

Measurement of the rotational kinematics of the thigh during running can be done in two ways. Firstly, movement of the thigh can be measured with respect to
an external reference (vertical) to obtain a segment angle. This angle has been commonly referred to as a thigh angle in the literature and has predominantly been measured in studies using 2D analysis [57,58,60–63,69,71–80]. Secondly, the movement of the thigh can be measured with respect to the proximal adjacent body segment to obtain a hip joint angle. The proximal adjacent body segment typically used is the pelvis [51,81–90], however, the trunk has been used on occasions [91,92]. Both 2D [82–84,86,87,90–92] and 3D [51,81,85,88,89] analyses have been used to measure the hip joint angle during running.

As a result of the increasing use of 3D analysis in biomechanical research, this review has been set out to conform with 3D terminology. During running, motion of the trunk segment, the pelvic segment and the thigh segment or hip joint can be described by rotations about three orthogonal axes. The rotations about the medial–lateral (M–L), anterior–posterior (A–P) and vertical axes can be seen as angular movement occurring in the sagittal, coronal and transverse planes respectively. As 3D motion is difficult to visualise, this review will discuss the three rotations separately. This layout is used for clarity and is not meant to detract from the fact that the rotations about the three axes occur simultaneously during running.

Studies investigating running at a variety of speeds have been included in this review. A discussion of speed-related effects will be made, where possible, for each of the rotations. However, one must bear in mind that, due to obvious methodological difficulties, there is limited information available at speeds representing the sprinting action. The precise effect of increasing speed-related effects will be made, where possible, for running speeds of 2–7 m/s, this mean angle ranges from 2.4 to 13° of trunk flexion (Table 2).

Several authors have found that during running the position of minimal trunk flexion tends to occur at or just prior to foot strike. The trunk then flexes during stance, with maximal trunk flexion occurring around mid to late stance [50,57–63,65]. The data from various authors for the degree of trunk flexion at foot strike, toe off, maximum hip extension and maximum hip flexion are illustrated graphically in Fig. 2. At slow running speeds the position of minimal trunk flexion occurs around foot strike. As speed increases, the position of minimal trunk flexion occurs during the airborne period preceding stance, such that by foot strike the trunk has already commenced flexion [50,65].

2.1. Rotation in the sagittal plane about a M–L axis

2.1.1. Trunk

Rotation of the trunk about a M–L axis during running occurs in the sagittal plane and is commonly known as trunk flexion and extension. All studies have considered the trunk to be a single segment and have measured with respect to an external reference (vertical). The total range of movement (amplitude) varies from 2.3 to 23° for speeds of 2–7 m/s (Table 1). As can be seen in Table 1, there does not appear to be a clear relationship between amplitude and running speed [50,60,61,65].

Rotation of the trunk about a M–L axis (flexion–extension) during running displays a regular pattern of two full oscillations per running cycle [50,58,59,65] (Fig. 1). The point about which this movement occurs is essentially the mean angle of trunk rotation about an M–L axis (average degree of trunk inclination with respect to the vertical) over the running cycle. For

<table>
<thead>
<tr>
<th>Reference</th>
<th>Analysis</th>
<th>Speed (m/s)</th>
<th>Amplitude (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carlson et al.</td>
<td>2D</td>
<td>2.5</td>
<td>5.0</td>
</tr>
<tr>
<td>Elliot and Blanksby [60]</td>
<td>2D</td>
<td>2.5–3.5</td>
<td>2.5–3.7</td>
</tr>
<tr>
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<td>2D</td>
<td>2.5–3.5</td>
<td>2.3–12.2</td>
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<td>Elliot and Roberts [62]</td>
<td>2D</td>
<td>5.2</td>
<td>5.8</td>
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<td>Elliot and Ackland [63]</td>
<td>2D</td>
<td>5.5</td>
<td>23.0</td>
</tr>
<tr>
<td>Thorstensson et al. [65]</td>
<td>2D</td>
<td>2.0–6.0</td>
<td>2.0–12.0</td>
</tr>
<tr>
<td>Thorstensson et al. [50]</td>
<td>2D</td>
<td>2.0–6.0</td>
<td>2.5–5.0</td>
</tr>
<tr>
<td>Bates et al. [57]</td>
<td>2D</td>
<td>7.4</td>
<td>3.0</td>
</tr>
</tbody>
</table>
The pelvis then anteriorly tilts, reaching a position of maximum anterior tilt around toe off. The pelvis then posteriorly tilts slightly during initial swing before anteriorly tilting again during terminal swing [51,88]. This second posterior and anterior tilt during swing is produced by the stance phase forces of the contralateral lower limb. The pattern of A–P tilting of the pelvis is similar at all speeds. Likewise, the amplitude of A–P tilting appears to increase very little with faster running velocities. It is thought that A–P tilting of the pelvis needs to be minimised to conserve energy and maintain efficiency in running [88].

2.1.3. Hip

When assessing the abundance of literature dealing with hip rotation about a M–L axis (flexion–extension) during running, it is important to note whether the thigh has been measured with respect to the vertical to obtain a segment angle (thigh angle) or has been measured with respect to the proximal adjacent body segment to obtain a joint angle (hip angle). As the trunk is generally inclined forward during running, studies measuring the hip angle will tend to show increased flexion angles and decreased extension angles with respect to studies measuring the thigh angle [76]. This is evident when comparing the data in Tables 4 and 5. Considerable variation in the results can also be seen to occur between 2D and 3D studies due to inherent differences in modeling procedures, as previously discussed. This is evident when comparing the data in Tables 5 and 6. For example, at comparable running speeds, maximum

### Table 2

<table>
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<tr>
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</tr>
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<td>2.5–3.5</td>
<td>6.8–7.6</td>
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<tr>
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<td>2.5–3.5</td>
<td>7.3–9.3</td>
</tr>
<tr>
<td>Morgan et al. [64]</td>
<td>2D</td>
<td>3.3</td>
<td>9.3–9.5</td>
</tr>
<tr>
<td>Williams and Cavanagh [68]</td>
<td>3D</td>
<td>3.6</td>
<td>2.4–5.9</td>
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<td>Wank et al. [66]</td>
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<td>7.0</td>
</tr>
<tr>
<td>Williams et al. [69]</td>
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<td>10.2</td>
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<td>6.0</td>
<td>10.2</td>
</tr>
<tr>
<td>Thorstensson et al. [50]</td>
<td>2D</td>
<td>2.0–6.0</td>
<td>10–13b</td>
</tr>
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<td>3D</td>
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<td>8.0</td>
</tr>
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<td>Bates et al. [57]</td>
<td>2D</td>
<td>7.4</td>
<td>7.5</td>
</tr>
</tbody>
</table>

*a Positive is trunk flexion; zero is vertically aligned trunk segment except:
*b Where zero is adjusted to be neutral standing posture.
Fig. 2. Reported angles for trunk rotation about an M–L axis (flexion–extension) at the points of foot strike, toe off, maximum hip flexion and maximum hip extension during the running cycle.

Upon viewing Table 6 it is apparent that differences exist between the results of studies using 3D analysis.

The hip flexion angles reported by Ounpuu [51] and Novacheck [88] differ from those of Pink et al. [89]. This discrepancy may be explained by variations in the positioning of markers to measure hip motion. Ounpuu [51] placed a rigid cluster of markers on the lateral thigh and placed markers on the right and left anterior...
 superior iliac spines and the mid-point between the posterior superior iliac spines for the pelvis. This protocol has been well described by Davis et al. [52]. Novacheck [88] used the standard marker set of the VICON motion analysis system (Oxford Metrics, Ltd., Oxford, UK), as reported in a previous publication by the same author [96]. This marker set is similar to that described by Davis et al. [52]. In contrast, Pink et al. [89] placed markers on the greater trochanter, the lateral trunk just above the iliac crest and the acromion.

The pattern of hip rotation about a M–L axis during running is illustrated in Fig. 1. At foot strike the hip is flexed (Tables 4–6). Differing results exist in the literature concerning the change in hip joint position during the absorption period of stance. Slight hip flexion has been found to occur following foot strike, often only at slower running speeds [58,76,82–84,86,87,89,97–99]. The slight hip flexion occurring at this time is believed to help absorb the increased impact forces of initial ground contact [99,100]. In contrast, others have found the hip to remain stationary momentarily during the absorption period [71,72,79,80]. Finally, some authors have found the hip to continue to extend during the absorption period of stance [51,75,78,88]. These differences between studies may be due to the variety of data acquisition speeds utilised, which range from 60 to 200 Hz, and variations in the measurement methods. They may also simply be due to differences in running techniques between subjects, as found by Cavanagh [72]. Finally, the discrepancies may well be speed related, as will be discussed later.

Hip extension then occurs through the middle and later stages of stance such that, by the time of toe off, the hip is in an extended position (Tables 4–6). It is evident upon viewing the reported hip positions at toe off that variation exists between studies. As explained earlier and as is apparent upon inspection of Tables 4–6, most of this variation may be attributable to the measurement of hip angles versus thigh angles and the use of 2D analysis versus 3D analysis. The position of maximal hip extension usually occurs at or just after toe off (Tables 4–6). Increases in the degree of hip extension at this time have been correlated with longer stride lengths [101].

During initial swing after maximum extension is obtained, the hip reverses direction and begins to rapidly flex (Tables 4–6). Dillman [100], Nilsson et al. [87] and Pink et al. [89] described the onset of hip flexion to occur close to the time of contralateral foot strike. However, other authors have found contralateral foot strike to occur well after the onset of hip flexion.

### Table 3

<table>
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<th>Speed (m/s)</th>
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<td>Cairns et al. [81]</td>
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### Table 4

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<th>Maximum flexion</th>
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<td>29</td>
<td>−29</td>
<td>−32</td>
<td>NR</td>
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* Negative value indicates hip flexion with respect to the vertical.
Table 5
Rotation of the thigh about an M–L axis (flexion–extension) during running with respect to the proximal body segment (°) (hip angle), from 2D measurementsa

<table>
<thead>
<tr>
<th>Reference</th>
<th>Speed (m/s)</th>
<th>Foot strike</th>
<th>Toe off</th>
<th>Maximum extension</th>
<th>Maximum flexion</th>
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<td>NR</td>
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<td>9.8</td>
<td>NR</td>
<td>−10−13</td>
<td>−12−22</td>
<td>55−59</td>
</tr>
<tr>
<td>Mann and Hagy [83]</td>
<td>10.8</td>
<td>48</td>
<td>18</td>
<td>18</td>
<td>96</td>
</tr>
<tr>
<td>Moran et al. [86]</td>
<td>10.8</td>
<td>48</td>
<td>18</td>
<td>18</td>
<td>96</td>
</tr>
</tbody>
</table>

a Negative value indicates hip extension with respect to the vertical.

[58,71,75,102]. Hip flexion continues through the first two thirds of swing phase until a maximal position is reached. Maximal hip flexion occurs at approximately the same time as contralateral toe off [44,58,71,75,87,102–105].

During the final third of swing the hip reverses direction again and begins to extend prior to foot strike (Tables 4–6). Dillman [106] demonstrated that this hip extension during the final stages of terminal swing reduces the horizontal velocity of the foot prior to foot strike and theorised that this would then help to minimise the retarding ground reaction force. Other investigators have also emphasised this possible function of terminal swing hip extension [51,75,88,90].

As running speed increases, the relative duration of the hip flexion phase of the running cycle increases [107] and the relative duration of the hip extension phase decreases [87,107]. The angle of hip flexion at foot strike demonstrates only small increases with faster running velocities [75,78,82–84,86,89,93,97–99,108]. Based on his comprehensive review in 1985, Williams [79] concluded that the angle of hip flexion at foot strike did not appear to change appreciably with increased running speed, at least at speeds above 4 m/s. Data listed in Tables 4 and 5 certainly supports this conclusion. Several authors have found hip flexion to increase slightly during the absorption period of stance at slow speeds of running only. At faster running speeds, the hip continues to extend through the absorption period [58,76,82–84,86,89,97–99]. However, other researchers have not demonstrated such an effect at their respective slow test speeds [75,78,88].

At the end of stance, maximal hip extension has been found to increase when running increases from slow to moderate speeds [75,76,78,82,84,86–90,97–99,109]. The magnitude of this increase is often only slight [78,87,88,90]. Sinning and Forsyth [90] believed that this is a reflection of how effectively the anterior capsule limits extension of the thigh at the hip joint. Williams [79] also suggested that anatomical features limit hip extension, as it appeared to him that maximal values of hip extension during running approached the limit for passive range of movement. Interestingly, Mann et al. [83,84,99] and Moran et al. [86] have found that with fast running the degree of maximal hip extension, whilst still occurring at toe off, is actually less than that which occurs at moderate speeds of running. This is supported by Saito et al. [107], Kunz and Kaufmann [110], Mann et al. [91] and Mann and Herman [92] who all found elite sprinters to have less hip extension at toe off in a bid to quickly begin recovery. Data in Table 4 also supports this effect, as the magnitude of hip extension can be seen to decrease above speeds of 6.5 m/s.

When comparing data from various authors, the slight differences in the timing of maximal hip extension with respect to toe off do not appear to be clearly speed related. Pink et al. [89] and Miller [75] found it to remain close to toe off regardless of speed. Novacheck [88], Mann et al. [84], Mann [99] and Moran et al. [86] all found maximal hip extension to occur during initial swing at slower speeds and at toe off at faster speeds. In contrast, Mann and Hagy [83] and Sykes [78] found maximal hip extension to occur slightly later in initial swing with increasing speeds.

The degree of hip flexion during swing phase shows the most consistent and dramatic increases with faster running. Increased hip flexion at this time is mainly responsible for the increase in the net amplitude of hip angular displacement evident with faster running speeds.
Table 6
Rotation of the thigh about an M–L axis (flexion–extension) during running with respect to the proximal adjacent body segment (°) (hip angle), from 3D measurementsa

<table>
<thead>
<tr>
<th>Reference</th>
<th>Speed (m/s)</th>
<th>Foot strike</th>
<th>Toe off</th>
<th>Maximum extension</th>
<th>Maximum flexion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ounpuu [51]</td>
<td>2.2</td>
<td>47</td>
<td>4</td>
<td>4</td>
<td>50</td>
</tr>
<tr>
<td>Pink et al. [89]</td>
<td>3.0</td>
<td>18</td>
<td>−6</td>
<td>−6</td>
<td>29</td>
</tr>
<tr>
<td>Novacheck [88]</td>
<td>3.2</td>
<td>50</td>
<td>−2</td>
<td>−4</td>
<td>57</td>
</tr>
<tr>
<td>Cairns et al. [81]</td>
<td>3.6</td>
<td>NR</td>
<td>NR</td>
<td>−9</td>
<td>53</td>
</tr>
<tr>
<td>Novacheck [88]</td>
<td>3.8</td>
<td>48</td>
<td>−6</td>
<td>−8</td>
<td>65</td>
</tr>
<tr>
<td>Pink et al. [89]</td>
<td>3.9</td>
<td>20</td>
<td>−11</td>
<td>−11</td>
<td>31</td>
</tr>
</tbody>
</table>

*a Negative value indicates hip extension with respect to the vertical.

(Tables 4–6). As a result, flexion of the hip during the first half of swing seems to play an extremely important role when increasing running speed [83,84,86]. It may be that greater hip flexion increases running speed by producing longer step lengths [88]. Deshon and Nelson [111] found increased hip flexion during swing to be significantly correlated with increased stride length.

2.1.4. Coordination of the M–L axis rotations

Summarising the results of past research, during running, the mean angle of trunk flexion is between 3 and 13° whilst the mean angle of anterior tilt of the pelvis is between 15 and 20° (Fig. 1). Novacheck [88] believed that these mean angles of the trunk and pelvis assisted in directing the resultant ground reaction force vector anteriorly during stance, facilitating forward acceleration.

The trunk and pelvis can then be seen to undergo two full oscillations about these mean angles per running cycle, whilst the hips flex and extend reciprocally (Fig. 1). At foot strike, the trunk is close to its position of minimal flexion, the pelvis is posteriorly tilting slightly and the stance hip is flexed. During stance, the hip extends as the trunk flexes and the pelvis begins to anteriorly tilt. By the end of stance, the trunk reaches its position of maximum flexion. At or just after toe off the pelvis reaches its position of maximum anterior tilt as the hip approaches maximum extension. During the airborne phase flexion of the trunk decreases, the pelvis begins to posteriorly tilt again and the hip on the swing side commences rapid hip flexion. Soon after, contralateral foot strike occurs (50% running cycle) and similar trunk and pelvis movements can be seen to occur with respect to contralateral stance.

It appears that rotation of the trunk segment, the pelvic segment and the thigh segment or hip joint about a M–L axis during running is highly coordinated. For example, the mean pelvic A–P tilt angle over the running cycle has been shown to have a significant negative correlation with maximum hip extension range of movement during running [112]. The mean angle of pelvic A–P tilt was found to be increased (i.e. more anteriorly tilted) in runners who displayed reduced hip extension during terminal stance. It may be that runners can increase their overall mean pelvic A–P tilt angle as a compensatory mechanism for a restriction in the available range of hip extension during terminal stance. It would be interesting to know whether this is also associated with an increase in the lumbar lordosis.

One group of authors [44,103–105] have described the lumbar spine as the pivotal point of the lower extremity lever system during running. Movement of the lower limb backwards during stance is believed to commence with extension of the lumbar spine, which then anteriorly tilts the pelvis to effectively increase the working range of extension and contribute to the extensor thrust mechanism of the lower limb. In order to allow the lumbar spine to participate in this proposed coordinated mechanism of the lower extremity, these authors felt that the trunk should essentially be in an erect position during the early stages of stance [44,103–105].

In support of this hypothesis, it has been shown that the most erect position of the trunk occurs close to foot strike [50,57–63,65]. However, in opposition, Sinning and Forsyth [90] questioned this role of the lumbar spine during running. As maximal hip extension was found to occur well after the completion of the extension movements at the primary propulsive joints (i.e. the ankle and the knee), it was suggested that there appeared to be little need for the lumbar spine to be involved in the proposed extensor thrust mechanism [90]. Therefore, movement of the trunk or, more specifically, the lumbar spine may be highly coordinated with movement at the hip during the propulsion period of stance, but this does not necessarily mean that it is playing a significant role in producing the propulsive forces.

2.2. Rotation in the coronal plane about an A–P axis

2.2.1. Trunk

Rotation of the trunk about an A–P axis (lateral flexion) during running has been measured using 2D analysis (Fig. 3). Thorstensson et al. [65] and Carlson et al. [59] placed skin markers over L3 and C7 and
measured the angle between the vertical and a line connecting these two markers (segment angle). Maximal angular displacement of the trunk to the left occurred during mid to early stance on the left side [59,65]. As running speed increased, the trunk reached its extreme lateral deviation to the left earlier, corresponding to the time of foot strike on the left [65]. Similar timing of the angular displacement of the trunk to the right occurred with respect to foot strike on the right side [59,65]. Thorstensson et al. [65] found the net amplitudes of the angular displacements to range from 4 to 14°, with no clear change in net values with increasing running speed. Carlson et al. [59] found a net amplitude of 7° at a speed of 2.5 m/s.

In an extension of their earlier work, Thorstensson et al. [50] placed markers on C7, L3 and midway between these two positions (50% markers). Two measures of trunk lateral flexion were then taken. Firstly, the angle between the vertical and a line connecting the C7 and the 50% markers was measured (segment angle). Secondly, the angle between a line connecting the C7 and 50% markers and a line connecting the L3 and 50% markers was measured (internal bending of the trunk). Maximal internal bending of the trunk to the left occurred just prior to left foot strike, whereas maximal overall tilting to the left of the trunk segment with respect to the vertical occurred during early stance on the left. The trunk then moved towards the right, with these angles reaching a maximum at the same time with respect to stance on the right. The timing of these movements was not found to change with increasing speeds of running. However, unlike Thorstensson et al. [65], as running speed increased from 2 to 6 m/s, the net angular displacements of the internal bending of the trunk and the overall tilting of the trunk were found to increase from 11 to 20° and from 5 to 11°, respectively [50].

### 2.2.2. Pelvis

Rotation of the pelvis about an A–P axis during running occurs in the coronal plane and is commonly known as pelvic obliquity or lateral pelvic tilt [51,81,88,94,104,105,113]. Variation exists between authors regarding the reported amplitude of pelvic obliquity during running (Table 7). This is more than likely

<table>
<thead>
<tr>
<th>Reference</th>
<th>Analysis</th>
<th>Speed (m/s)</th>
<th>Amplitude (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ounpuu [51]</td>
<td>3D</td>
<td>2.2</td>
<td>2.0</td>
</tr>
<tr>
<td>Cairns et al. [81]</td>
<td>3D</td>
<td>3.6</td>
<td>14.9</td>
</tr>
<tr>
<td>Novacheck [88]</td>
<td>3D</td>
<td>3.2–3.8</td>
<td>7.0–12.0</td>
</tr>
</tbody>
</table>
due to differences in subjects and running velocities between studies. Cairns et al. [81], testing a group of ten racewalkers running at 3.6 m/s, reported an amplitude of 15° of pelvic obliquity. It is possible that racewalkers, when running, carry over an exaggerated pelvic movement pattern to some extent in comparison to actual runners. In contrast, Ounpuu [51,114] reported an amplitude of 2°. This small value may be attributable to the slower running velocity (2.23 m/s) or the young age of the subjects (5–11 years).

The pattern of pelvic rotation about an A–P axis during running is shown in Fig. 3. At foot strike the pelvis is obliquely aligned, being slightly higher on the stance (ipsilateral) side and slightly lower on the swing (contralateral) side [88]. Pelvic obliquity is thought to play a role in shock absorption and in controlling the smooth descent and ascent of the body’s centre of gravity at this time [104,105]. By midstance though, the pelvis becomes horizontal. The pelvis then continues to elevate on the swing (contralateral) side, reaching a maximum downward obliquity on the stance (ipsilateral) side around toe off. During the airborne period, the pelvis then begins to rise on the initial swing (ipsilateral) side and lower on the terminal swing (contralateral) side as it approaches foot strike. This movement arrests and reverses direction slightly with foot strike of the contralateral extremity. During the remaining mid and terminal ipsilateral swing though, the pelvis then continues to rise on this side until maximum upward obliquity is reached during terminal swing. The elevation of the swing side pelvis during terminal swing maintains foot clearance as the swing side hip and knee are extending [88].

2.2.3. Hip

Numerous studies have investigated the rotation of the hip about an A–P axis (abduction–adduction) during running with remarkably consistent results (Table 8). All studies have measured the motion of the thigh segment with respect to the pelvis to obtain a hip joint angle. The pattern of motion is illustrated graphically in Fig. 3. At foot strike, the hip is in an adducted position. During initial stance, when shock absorption is occurring, hip adduction increases slightly [51,82,88,97,99,105]. From midstance until toe off, during the propulsive period of stance, the hip progressively abducts reaching a slightly abducted position by toe off. Further hip abduction continues during the first half of swing. Maximal hip abduction of the swing extremity occurs around midswing. During terminal swing, the hip begins to adduct again [51,82,88,97,99]. The hip abduction during terminal stance and early swing is thought to aid clearance of the contralateral swinging limb. The hip adduction during terminal swing possibly functions to accurately position the lower limb in preparation for a stable initial contact [51].

2.2.4. Coordination of the A–P axis rotations

Based on the research performed to date, rotation of the trunk segment, the pelvic segment and the thigh segment or hip joint about an A–P axis during running also appears to be intimately coordinated (Fig. 3). At foot strike on the left (for example), the trunk is laterally flexed towards the left, the pelvis is upwardly oblique on the left side (i.e. lower on the swing side) and the left hip is adducted. During early stance, left hip adduction is seen to increase slightly as a shock absorbing mechanism, and this appears to be closely coupled to an increase in the lateral flexion of the trunk towards the left. An associated increase in the upward obliquity of the pelvis on the left side at this time, however, does not seem to occur. It may be that lateral translation of the pelvis, rather than obliquity, is actually producing the increase in hip adduction during early stance. After early stance on the left, the trunk begins to laterally flex towards the right and the pelvis begins to elevate on the contralateral (swing) side such that by terminal stance, the trunk is laterally flexed to the right, the contralateral (swing) side of the pelvis is elevated and the left hip is abducted (Fig. 3).

The trunk and pelvic movements are coordinated with the stance phase of the right extremity during the swing phase of the left extremity. The trunk is laterally flexed to the right and the pelvis is downwardly oblique on the left side (upwardly oblique with respect to the right side) during initial swing, whilst during terminal swing the trunk is beginning to laterally flex towards the left and the pelvis is rising on the left side (becoming upwardly oblique on the left). The left hip is also adducting slightly at this time in preparation for foot strike (Fig. 3).

The combined rotational movements occurring at the lumbar spine and hip joints in the coronal plane are thought to play a vital role in decoupling the intense lower extremity motion from the shoulder and head [88]. It is believed that this then minimises head and shoulder motion, allowing lateral balance and equilibrium to be maintained [88,104]. Further research is
Table 9
Amplitude of pelvic rotation about a vertical axis (axial rotation) during running

<table>
<thead>
<tr>
<th>Reference</th>
<th>Analysis</th>
<th>Speed (m/s)</th>
<th>Amplitude (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ounpuu [51]</td>
<td>3D</td>
<td>2.2</td>
<td>16.0</td>
</tr>
<tr>
<td>Cairns et al. [81]</td>
<td>3D</td>
<td>3.6</td>
<td>16.8</td>
</tr>
<tr>
<td>Novacheck [88]</td>
<td>3D</td>
<td>3.2–3.8</td>
<td>16.0–18.0</td>
</tr>
</tbody>
</table>

required, detailing the precise nature of the coodination within the lumbo–pelvic–hip complex in the coronal plane during running, in order to determine whether this is indeed true.

2.3. Rotation in the transverse plane about a vertical axis

2.3.1. Trunk

It is thought that the lower spine rotates backward with extension of the trailing leg while, at the same time, the upper spine rotates forward synchronously with the arm on the same side to maintain equilibrium during running [105,115]. To date this movement has proven to be extremely difficult to accurately measure and consequently there are no published studies that quantify the degree of rotation of the lumbar spine or trunk about a vertical axis during running. 3D analysis has been used to successfully quantify rotation of the lumbar and/or thoracic spine about a vertical axis during walking [34–36,43,116–122]. Hopefully, as improvements in 3D modeling procedures of the spine occur, it will become possible to quantify this motion during running.

2.3.2. Pelvis

Rotation of the pelvis about a vertical axis during running occurs in the transverse plane and is commonly known as axial rotation or internal and external rotation [51,81,88,94,97–99,104]. Internal pelvic rotation is when the reference side of the pelvis is anterior and external pelvic rotation is when the reference side of the pelvis is posterior. The amplitude of this movement is between 16 and 18° (Table 9).

Previous qualitative descriptions of pelvic rotation about a vertical axis during running describe the movement pattern to be similar to that of walking [45,97–99,104,105]. However, interestingly, 3D quantitative analyses have shown the pattern during running to be opposite to that which occurs during walking [51,88]. The pelvis is internally rotated during initial stance in walking in order to increase stride length, whilst during running it appears that the pelvis is no longer utilised as a stride lengthening mechanism.

![Fig. 4. Rotation of the pelvis and hip about a vertical axis during running. Pelvic internal rotation: reference side of the pelvis is anterior with respect to the contralateral side. Pelvic external rotation: reference side of the pelvis is posterior with respect to the contralateral side. Based on data from: Novacheck [88].](image-url)
The pattern of pelvic rotation about a vertical axis during running is illustrated in Fig. 4. The pelvis externally rotates on the side of the lower limb preparing for foot strike. By foot strike the pelvis is slightly externally rotated on the stance side, which continues to increase until a maximal position of external rotation is reached around midstance. During terminal stance the pelvis begins to internally rotate on the stance side such that by toe off the pelvis is in a neutral position. Internal rotation of the pelvis on the swing side continues through the early swing period, reaching a maximal position of internal rotation around midswing. The pelvis then begins to externally rotate again on the side of the lower limb in terminal swing [88]. A similar pattern of movement has been reported by Ounpuu [51]. In contrast to Novacheck [88], Ounpuu [51] found neutral pelvic rotation to occur prior to toe off, during late stance. This difference is likely to be due to the considerably slow running speed of the subjects in Ounpuu’s study [51,114].

It is believed that the pattern of pelvic rotation about a vertical axis during running is important for energy efficiency [88]. For example, an externally rotated pelvis on the side of the lower limb preparing for foot strike may actually assist in decreasing the posterior component of the ground reaction force. Relative to an internally rotated pelvis, an externally rotated pelvis at foot strike decreases the horizontal linear distance between the point of initial contact and the body’s centre of gravity (Fig. 5). This may then minimise the horizontal braking forces and aid in avoiding potential loss of speed. Further research is required to validate this hypothesis.

2.3.3. Hip

Numerous authors have provided qualitative descriptions of the vertical axis rotation of the hip joint during running [97–99,104,105,123]. Mann [97,99] based his observations on the qualitative analysis of high-speed films of running which he reported to demonstrate the same basic type of rotation as occurs with walking. At foot strike the hip joint is thought to be externally rotated [104,105,123]. During the absorption period of stance, the hip joint then internally rotates [97–
returning to a neutrally rotated position by terminal crease in internal rotation during mid swing before rotated position, whilst Novacheck [88] found an in-
onpuu [51] showed the hip to remain in a neutrally rotated position, whilst Novacheck [88] found an increase in internal rotation during mid swing before returning to a neutrally rotated position by terminal swing.

2.3.4. Coordination of the vertical axis rotations

The precise nature of the coordination of the vertical axis rotations within the lumbo–pelvic–hip complex is largely unknown. What has been shown from 3D analyses to date is that the pelvis is slightly externally rotated on the stance side at foot strike. During the absorption period of stance the pelvis continues to externally rotate on the stance side as the hip joint internally rotates. Maximal pelvic external rotation and hip internal rotation occur at midstance. Because hip rotation is defined as movement of the thigh segment relative to the pelvic segment, it is likely that the internal hip rotation at this stage is predominantly a consequence of the externally rotated pelvic segment. The thigh segment in absolute terms may actually appear straight to the observer. During the propulsive period of stance the pelvis then internally rotates on the stance side whilst the hip joint externally rotates to approximately neutral position [51,88] (Fig. 4).

Mann [97–99] hypothesised that the external rotation of the hip during terminal stance is an active process brought about by the forward motion of the swing side of the pelvis. He believed that as the pelvis on the swing side is moved forward by the rapidly flexing thigh, it functions as a crank to impart external rotation to the stance limb [97–99]. However, this hypothesis is based on the assumption that the pelvis is rotating forward on the swing side, as occurs during walking. As noted, recent 3D studies [51,88] have indicated that this is not the case for running.

In an attempt to explain how the spine contributes to human locomotion, Gracovetsky and associates [45–47] developed the theory of the ‘spinal engine’. The theory explains how the lordotic lumbar spine converts a lateral flexion movement (rotation of the lumbar spine about a sagittal axis) into an axial torque, which rotates the pelvis about a vertical axis. Lateral flexion of the lumbar spine towards the left, for example, is described to induce a clockwise (from above) rotation of the pelvis. Gracovetsky et al. [41,45,47–49] then applied the theory to running as follows. The hip extensors contract during late stance to lift the body in the gravitational field. In the flight phase after toe off, the unloaded spine fires its erector spinae to produce lateral flexion of the spine towards the side of the extremity approaching foot strike. Lateral flexion of the spine towards the stance side is further increased at foot strike due to the ground reaction forces. The coupled motion of the spine transforms this lateral flexion into an axial torque that rotates the pelvis about its vertical axis.

Gracovetsky [48] demonstrated that the ‘spinal engine’ theory is capable of successfully explaining in rational form a considerable body of experimental facts. For example, kinematic data of the lumbar spine and pelvis during walking provided evidence for the theory. However, pelvic rotation about a vertical axis during running, as previously discussed, has been found to be opposite in direction to that of walking. This data conflicts with the predictions of the ‘spinal engine’ theory for running. Gracovetsky himself points out that it is simply a question of time before new evidence forces a review of the theory. Recent data from Novacheck [88] and Ounpuu [51] regarding the rotation of the pelvis about a vertical axis during running suggests that the ‘spinal engine’ theory may need to be adjusted in terms of its applicability to running.

Based on his 3D data, Novacheck [88] described rotation of the pelvis about a vertical axis to function as a pivot between the forward swinging leg and the counter-rotating upper trunk and arms during running. At right toe off, during the start of double float, the right hip was maximally extended and the left hip was maximally flexed, whilst the right shoulder was rotated forwards and the left shoulder was rotated backwards.
3. Clinical implications

One of the basic objectives for studying movement patterns during running is to provide information that may assist in reducing susceptibility to injury. Running injuries can be the result of a mechanical overloading of the locomotor system (or part of it). Injuries due to overload are the result of forces acting on the human body [126]. During running, external forces act on the body, including ground reaction forces and inertial forces from moving body segments, which need to be balanced by internal forces developed from muscles, tendons, ligaments and joint capsules [127]. When these forces in a particular anatomical structure exceed critical limits, an injury may result. This occurs when one force is above a critical limit (e.g. acute injury) or when a number of cyclic forces are below that limit, but produce a combined fatigue effect (e.g. chronic overuse injury) [126].

It is logical to suggest that a particular kinematic pattern, demonstrated to be related to injury, is a reflection of forces that are exceeding critical limits. The question that then remains is whether a particular kinematic pattern during running is related to injury. In a review article discussing the role biomechanical research has played in reducing the frequency of sports injuries, Nigg [12] found sufficient evidence to suggest that correlations between running kinematics and injury can be found in certain circumstances. Perhaps most attention with regards to this has concentrated on establishing the connection between excessive pronation of the rearfoot and injury [19,20,22,24–29,128].

With regards to the lumbar spine, pelvis and hips, numerous authors have anecdotally reported relationships between disturbances to the normal kinematic pattern of these segments during running and injury. Increased lumbar lordosis and anterior pelvic tilt, thought to be associated with tightness of the hip flexor musculature, is an example of one abnormality that has been discussed [129,130]. Repetitive impingement of the vertebral facets from hyperextension of the lumbar spine has been suggested to be related to the onset of low back pain in runners [105,131] whilst increased anterior tilt of the pelvis during running has been cited as a predisposing factor for hamstring strains [129] [132]. Klein and Roberts [129] and Geraci [132] believed that such a position of the pelvis moved the ischial tuberosities superiorly, which altered the length of the hamstrings and caused premature fatigue. A greater hip flexion angle at foot strike has also been shown to be a factor that may lead to excessive stress being placed on the hamstrings [133]. Finally, increased pelvic rotation about an A–P axis (pelvic obliquity) during running has been implicated with iliobibial band friction syndrome [134] and sacroiliac joint injuries [135]. Wiklander and colleagues [136] reported a significant negative correlation between the degree of pelvic obliquity and the flexibility of the hamstrings. Interestingly, in comparison to sprinters, this group of investigators found that distance runners had stiffer hamstrings, greater pelvic obliquity and suffered more exertion injuries around the hips and lower back [3,136–138]. In order to verify any of these proposed relationships, an understanding of the normal kinematic coordination within the lumbo–pelvic–hip complex during running is clearly the crucial starting point.

4. Conclusion

This literature review has described the coordinate movement of the lumbo–pelvic–hip complex during running by combining the results of research performed to date. Many improvements in the understanding of the motion patterns in this area during running have occurred with the recent proliferation of commercially available 3D kinematic measurement systems. However, it is evident from the review that many deficiencies in the literature still exist. The 3D studies that have been performed to date have used either young children as subjects, small subject numbers or slow running speeds. This makes it extremely difficult to extrapolate the results to the normal population at risk of running injuries. In addition, there are no published data available that accurately details the 3D kinematics of the lumbar spine during running. Further research is therefore required to extend the current level of knowledge of the coordinate movement patterns within the lumbo–pelvic–hip complex during running. Future investigations need to consider the possible influence of variables such as age, gender and performance level of the subjects. This will no doubt provide valuable information to facilitate the understanding of the pathomechanics of this area.

References


