THREE-DIMENSIONAL MEASUREMENT OF REARFOOT MOTION DURING RUNNING

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Abstract—Excessive ranges of motion during running have been speculated to be connected to injuries to the lower extremities. Movement of the foot and lower leg has commonly been studied with two-dimensional techniques. However, differences in the alignment of the longitudinal axis of the foot with the camera axis will produce measurement errors for projected angles of the lower extremities. A three-dimensional approach would not have this limitation.

The purpose of this study is to present a three-dimensional model for calculation of angles between lower leg and foot, lower leg and ground, and foot and ground, and to compare results from treadmill running derived from this model with results derived from a two-dimensional model for different alignment angles between foot axis and camera axis.

It was found that several two-dimensional variables measured from a posterior view are very sensitive to the alignment angle between the foot and the camera axis. Some variables change as much as 1° for every 2° of change of the alignment angle. The large influence of rotations other than the measured one in two-dimensional measurements makes advisable the use of a three-dimensional model when studying motion between foot and lower leg during running.

INTRODUCTION

Several studies suggest that a correlation exists between excessive pronation and various injuries of the lower extremities (Clarke et al., 1983; Frederick, 1986; Nigg and Baehlsen, 1988). A prospective epidemiological study (Luethi et al., 1986) showed that the angle between the lower leg and the rearfoot measured in the frontal plane (which was used as an indicator of eversion) was a relevant variable predicting the occurrence of pain and/or injuries in the lower extremities of tennis players. Baehlsen (1988) showed in a prospective epidemiological study that excessive pronation is a good predictor of various overuse injuries of runners, specifically of the patella-femoral syndrome.

Studies analysing rearfoot and lower leg motion have been concentrating on two-dimensional motion analysis methods (Bates et al., 1978; Nigg and Morlock, 1987; Nigg et al., 1988; Stacoff et al., 1988; Vagenas and Hoshizaki, 1988). Two markers were attached to the posterior aspect of the lower leg indicating its longitudinal axis, and two markers to the heel cap of the shoe located so that the line between the markers form an angle of 90° to the horizontal in the unloaded shoe. The markers were recorded from a posterior view, projecting the markers into a plane perpendicular to the running direction. The two-dimensional Achilles tendon angle, β, was defined as the projection of the angle between the line combining the markers on the lower leg and the line combining the markers on the heel cap of the shoe; the rearfoot angle, γ, as the projection of the angle between the line on the shoe and the ground onto the film plane. Using these angle definitions, the variables commonly analysed include the initial Achilles tendon angle, β₀, the maximum Achilles tendon angle, βₓₓ, the total pronation, Δβₓₓ (β₀ - βₓₓ), the maximum pronation velocity, βₓₓ, the initial rearfoot angle, γ₀, the total pronation of rearfoot angle, Δγₓₓ, and the minimum rearfoot angle, γₘᵢₙ. Angles measured in a lateral view are defined in a similar manner.

The results of the two-dimensional technique are affected by projection errors, which depend on the alignment of the segments with the film plane. Moreover the two-dimensional measurement technique does not allow for calculation of the abduction angle between foot and lower leg which has been speculated to be of importance when searching for causes of injuries (Cavanagh, 1987; Engsberg and Andrews, 1987).

Only a few attempts have been made to analyse rearfoot and lower leg motion three-dimensionally (Soutas-Little et al., 1987; Engsberg and Andrews, 1987).

Description of three-dimensional motion between the foot and the lower leg can be made in at least three principally different ways. Firstly, the rotations can be measured about the ankle and subtalar joint if the positions of these axes are known. Secondly, the rotations can be measured about the clinical axes describing eversion/inversion, dorsi-/plantar flexion and ab-/adduction. Thirdly the motion can be described as rotation and translation about a screw axis, which does not have any anatomical correspondence.

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Soutas-Little et al. (1987) proposed a three-dimensional model for measurement of rearfoot motion angles based on the second approach using a system of Euler angles combined to a 'joint co-ordinate system'. The sequence of rotation can be chosen in several ways and the effect of different sequences was not investigated. Furthermore, their model is restricted to calculations of motion between the segments, and leaves angles between a segment and the ground undefined.

The purpose of this paper is:
(a) to provide a three-dimensional model for calculation of the angles between the foot and the lower leg and between the ground and the segments;
(b) to apply the model to an experimental situation and calculate the reliability of measured experimental variables;
(c) to compare the results derived from this three-dimensional model with results derived from the two-dimensional model for different camera/foot alignments.

METHOD

Glossary

\( \tilde{X}, \tilde{Y} \) and \( \tilde{Z} \) orthogonal coordinate axes fixed in space
\( \tilde{I}, \tilde{J}, \) and \( \tilde{K} \) orthogonal coordinate axes fixed in the rearfoot
\( \tilde{I}, \tilde{J}, \) and \( \tilde{K} \) orthogonal coordinate axes fixed in the lower leg
\( A, B \) and \( C \) the positions of three markers on the heel-cap of the shoe expressed in the coordinates of the \( \tilde{X} \tilde{Y} \tilde{Z} \) system
\( D, E \) and \( F \) the positions of three markers on the posterior aspect of the lower leg expressed in the coordinates of the \( \tilde{X} \tilde{Y} \tilde{Z} \) system

Three orthogonal coordinate systems are defined. The Room Coordinate System, RCS, is right-handed and fixed in space with the \( \tilde{Y} \)-axis in the running direction and the \( \tilde{Z} \)-axis perpendicular to the floor. The Leg Coordinate System, LCS, is right-handed in the right leg and left-handed in the left leg and is fixed to the lower leg with the \( I_t \)-axis parallel to the longitudinal axis of the lower leg and the \( J_t \)-axis in the lateral direction. The Foot Coordinate System, FCS, is right-handed in the right foot and left-handed in the left foot and is fixed in the foot, with the \( J_t \)-axis parallel to the longitudinal axis of the foot and the \( K_t \)-axis being perpendicular to the shoe sole of the rearfoot (Fig. 1). The directions of the axes are defined at a standardized standing posture ensured by a calibration fixture described below.

The three-dimensional model consists of two parts. The first part includes calculation of angles between the RCS and each of the segment coordinate systems. Based on the two-dimensional definitions proposed by Nigg (1986) the following three-dimensional definitions have been made:

- \( \gamma \) rearfoot angle. The angle between \( K_t \) and the \( \tilde{X} \tilde{Y} \) plane of the RCS measured in the \( \tilde{I}_r \tilde{K}_r \) plane on the medial side. \( \gamma = \arccos(\tilde{K}_r \cdot (\tilde{Z} \times \tilde{J}_r)) \)
- \( \alpha \) angle of the lower leg. The angle between \( K_t \) and the \( \tilde{X} \tilde{Y} \) plane measured in the \( \tilde{I}_r \tilde{K}_r \) plane on the medial side. \( \alpha = \arccos(\tilde{K}_r \cdot (\tilde{Z} \times \tilde{J}_r)) \)
- \( \delta \) flexion angle of the rearfoot. The angle between \( K_t \) and the \( \tilde{X} \tilde{Y} \) plane measured in the \( \tilde{J}_r \tilde{K}_r \) plane posteriorly. \( \delta = \arccos(\tilde{K}_r \cdot (\tilde{I}_r \times \tilde{Z})) \)
- \( \eta \) flexion angle of the lower leg. The angle between \( K_t \) and the \( \tilde{X} \tilde{Y} \) plane measured in the \( \tilde{J}_r \tilde{K}_r \) plane posteriorly. \( \eta = \arccos(\tilde{K}_r \cdot (\tilde{I}_r \times \tilde{Z})) \)
- \( i \) abduction of the rearfoot. The angle between \( J_t \) and the \( \tilde{Y} \tilde{Z} \) plane measured in the \( \tilde{J}_r \tilde{X} \) plane. \( i = \arccos(\tilde{J}_r \cdot \tilde{X}) \)

Fig. 1. Orientation of coordinate systems and definition of some angles.
κ abduction angle of the lower leg. The angle between $\hat{J}_1$ and the $\hat{YZ}$ plane measured in the $\hat{X} \hat{Y} \hat{Z}$ plane. \( \kappa = 90^\circ - \arccos(\hat{J}_1 \cdot \hat{X}) \).

The second part of the model includes calculation of angles between the segments. The angles describing the movement between the foot and the lower leg can be defined in several ways. It seems appropriate to propose a model which corresponds to the terminology used by clinicians, namely eversion/inversion, dorsi-/plantar flexion and ab-/adduction.

It is proposed to first define dorsi-/plantar flexion, \( \zeta \), about the $\hat{I}_t$ axis since this axis approximates the ankle joint best. However, there is no anatomical preference in which order adduction/abduction and eversion/inversion angles, \( \theta \) and \( \beta \), should be calculated since these are not anatomical angles. The subtalar joint axis about which pronation does occur is oblique to these axes.

Results from both alternatives of sequence will be compared. The sequence 'flexion-eversion-abduction' will be referred to as FEA and corresponds to the solution suggested by Soutas-Little and co-workers (1987). The sequence 'flexion-abduction-eversion' will be referred to as FAE. Their mathematics are shown in Table I and illustrated in Fig. 2.

A two-dimensional analysis was made by projecting the $\hat{K}_t$ and $\hat{K}_f$ axes onto the $\hat{X} \hat{Y} \hat{Z}$ plane, which is perpendicular to the running direction, to obtain angular values from a posterior view, and by projecting the $\hat{J}_t$ and $\hat{K}_i$ axes onto the $\hat{X} \hat{Z}$ plane to obtain angular values from the lateral view.

Results were taken for a test subject, running on a treadmill at a speed of 3.8 m s\(^{-1}\) (Fig. 3). Three markers each were arbitrarily positioned on the heel-cap of the shoe (markers A, B and C) and on the lower leg (D, E and F). The markers were mounted on light, rigid plates in order to place the markers wider apart to increase the accuracy of the vector calculations.

A three-dimensional Setspot Multilab system was used to determine the position of the markers attached to the foot and the lower leg. The sampling frequency was 400 Hz filtered with a moving average filter with step 2 and length 4 reducing the frequency to 200 Hz. The stereo configuration of the two cameras used was calibrated with a calibration frame giving the orientation of the room coordinate system. The room coordinate system was aligned with the running direction of the treadmill. A calibration routine was developed to define the orientation of the segment coordinate systems FCS and LCS independent of the actual marker positions. A calibration fixture (Fig. 4) was developed to align the segment coordinate systems with the room coordinate system before or after each trial. It has two pairs of adjustable clamps which are placed on the malleoli and on the fibular head and the lateral condyle respectively. The clamps keep the midpoint between the fibular head and the lateral condyle right above the midpoint between the malleoli. The second metatarsal bone was placed under a guide line to indicate the longitudinal axis ($J_f$-direction) of the foot (Fig. 4).

At calibration, the foot and lower leg were put in the fixture, a picture was taken, and the orientations of the provisional coordinate systems (defined by the markers) were compared with the room coordinate system which was parallel to the desired segment systems.

<table>
<thead>
<tr>
<th>Alternative FEA</th>
<th>Alternative FAE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rot. axis</td>
<td>Equation</td>
</tr>
<tr>
<td>Dorsi-/plantar flexion</td>
<td>$\hat{I}_t$</td>
</tr>
<tr>
<td>Eversion/inversion</td>
<td>$\hat{I}_1 \times \hat{K}_f$</td>
</tr>
<tr>
<td>Abduction/adduction</td>
<td>$\hat{K}_i$</td>
</tr>
</tbody>
</table>

Fig. 2. Mechanical principles for alternatives FEA and FAE.
Fig. 3. The measurement set-up.

Fig. 4. The calibration fixture and the direction of the calibrated coordinate axes.

Fig. 5. Orientation of the provisional coordinate system of the rearfoot in relation to the markers.

During calibration, a rotational matrix which relates the provisional coordinate system with the segment system was calculated for each segment.

A provisional coordinate system associated with the positions of the markers on the right foot (see Fig. 5) is calculated as:

\[ \hat{I}_i = [(B - A)/(B - A)'] \]
\[ \hat{J}_i = (C - A) \times \hat{I}_i / (C - A) \times \hat{I}_i \]
\[ \hat{K}_i = \hat{I}_i \times \hat{J}_i. \]

With matrix formulation, the Provisional Coordinate system of the foot, \([PCF]\), is expressed as:

\[
[PCF] = \begin{bmatrix}
I_i & J_i & K_i
\end{bmatrix}
\]

The rotational Transformation Matrix of the Foot, \([TMF]\), is

\[
[TMF] = [PCF]^{-1}[RCS].
\]

The foot coordinate system, \([FCS]\), may then be calculated for each sample of the measurements by postmultiplying the transformation matrix, \([TMF]\), with the provisional coordinate system, \([PCF]\), of every sample.

\[
[FCS] = [PCF][TMF].
\]

In a similar way \([LCS]\) was determined for the lower leg:

\[
[LCS] = [PCI][TM1].
\]

The reliability of the system was tested by comparing the maximum eversion of the Achilles tendon angle, \(\beta_{max}\), of the right foot (a) for 50 consecutive footfalls; (b) for the average of 5-6 consecutive steps measured at 8 different occasions; (c) for the average of 5-6 consecutive steps measured at 5 different occasions with re-positioning and re-calibration of the cameras and re-calibration of the segment coordinate systems between each trial. The calculations were made with the result obtained with the two alternative
three-dimensional models as well as with the two-dimensional model using the same footfall material.

Different alignments of the longitudinal axis of the foot with the camera axis at two-dimensional measurements will produce different magnitudes of measurement errors. The sensitivity of the result for different foot/camera alignment angles was investigated by systematically changing the fictive camera position for a single representative footfall (Fig. 6). The interval from 10° of adduction to 30° of abduction of the foot compared with the camera axis was investigated. Both

![Fig. 6. Different alignment angles of the camera axis with respect to the longitudinal axis of the foot.](image)

<table>
<thead>
<tr>
<th></th>
<th>Two-dimensional $\beta_{\text{max}}$</th>
<th>Two-dimensional SD</th>
<th>Three-dimensional (FEA) $\beta_{\text{max}}$</th>
<th>Three-dimensional (FAE) $\beta_{\text{max}}$</th>
<th>Three-dimensional SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Consecutive footfalls</td>
<td>-6.7</td>
<td>0.80</td>
<td>-5.9</td>
<td>-5.9</td>
<td>0.74</td>
</tr>
<tr>
<td>Between occasions*</td>
<td>0.44</td>
<td>0.40</td>
<td>0.44</td>
<td>0.44</td>
<td>0.41</td>
</tr>
<tr>
<td>Between occasions*</td>
<td>0.54</td>
<td>0.44</td>
<td>0.44</td>
<td>0.44</td>
<td>0.46</td>
</tr>
</tbody>
</table>

*Each occasion is the average of five to six steps.

![Table 2. Reliability of the measurements](image)

![Fig. 7. Rearfoot angle, eversion angle and flexion angle as function of time for different alignment angles between foot and camera axis.](image)
Two-dimensional variables which have been used in several investigations were investigated for different foot/camera alignment angles and compared with the corresponding three-dimensional results.

**RESULTS AND DISCUSSION**

The difference between the maximum eversion of the Achilles tendon angle, $\beta_{\text{max}}$, obtained by the two alternative three-dimensional sequences proposed (FEA and FAE) was about 0.04° which is less than one-tenth of the standard deviation of consecutive footfalls (which was more than 0.7°). The small difference in the results is due to the small rotation about the subtalar joint axis during running.

The standard deviation between consecutive steps is about twice as large as between measurement occasions, where each occasion is the average of 5–6 steps. Re-calibration of the whole system increased the standard deviation only slightly (Table 2). As the same footfalls and the same markers are used at the calculations with each model (two-dimensional, FEA and FAE) the only source of variation in the result between them is their different definitions.

Figure 7 shows rearfoot angle, eversion angle, and dorsi-plantar flexion angle vs time, calculated with the two-dimensional and the three-dimensional models.

![Graph](image)

**Fig. 8.** The two- and three-dimensional values of (a) relative rearfoot angle, rearfoot angle at touch down and minimum rearfoot angle; (b) relative eversion angle, eversion angle at touch down and maximum eversion angle as function of the alignment angle between foot and camera axis.
for several different alignment angles between the foot and the camera axis. The two-dimensional results obtained from a posterior view (eversion angle, $\beta$, and rearfoot angle, $\gamma$) were strongly affected by the alignment of the foot with respect to the camera. The result of the two-dimensional rearfoot angle corresponded well with the three-dimensional result during the stance phase when the camera axis was aligned with the longitudinal axis of the foot. With greater abduction angle of the foot, a more pronated angle in the first part of the support phase and a more supinated angle in the last part of the support phase were obtained. The errors were smallest during mid-stance. None of the alignment angles provided a two-dimensional eversion angle that corresponded reasonably well with the three-dimensional result over the whole stance phase. In this case 10° of abduction of the foot gave the best correspondence during mid-stance but underestimated the angle during the initial part and overestimated it during the last part of the support phase.

The two-dimensional results obtained from a lateral view (the flexion angle, $\xi$) were not sensitive to the alignment angle, and corresponded well with the three-dimensional measurement results. This was expected, as this motion is mainly performed in the film plane. The result is in correspondence with the result of Soutas-Lit altercation (1987).

In Fig. 8a the variables $\gamma_0$, $\gamma_{\text{min}}$ and $\Delta\gamma_{\text{pro}}$ calculated with the two-dimensional and the three-dimensional models are plotted as a function of the alignment angle. Two-dimensional $\gamma_0$ and $\Delta\gamma_{\text{pro}}$ depended strongly upon the alignment angle, predicting a lower angular value the more abducted the foot was. A change of 2° in the alignment angle resulted in an approximate change of 1° in the computed angle values. The two-dimensional $\gamma_{\text{max}}$ was less sensitive. Still, a change of about 9° in the alignment angle resulted in a change of about 1° in $\gamma_{\text{min}}$. The two-dimensional values were close to the three-dimensional values only when the foot was well aligned with the camera axis.

The variables $\beta_0$, $\beta_{\text{max}}$ and $\Delta\beta_{\text{pro}}$ as a function of the alignment angle are shown in Fig. 8b. For the absolute values $\beta_{\text{max}}$ and $\beta_0$ the offset between two-dimensional and three-dimensional measurement values changes with the alignment angle. A change in alignment angle of about 2.7° results in a change in angle value of about 1°.

For the relative angle value $\Delta\beta_{\text{pro}}$ the offset is rather constant for different alignment angles. The variation in $\Delta\beta_{\text{pro}}$ over the whole interval of alignment angles investigated was less than 1.5°. It should be observed that $\beta_{\text{max}}$ and $\beta_0$, calculated with the two-dimensional method do not correspond best with the three-dimensional values when the alignment angle is close to zero as the rearfoot angle variables did.

The two-dimensional model is different from the three-dimensional model mainly in the sense that the angles are calculated about axes fixed in the laboratory instead of about axes following the movements of the segments. The three-dimensional model therefore provides angles that are anatomically more relevant. Most of the results obtained with the two-dimensional model cannot be compared with the results of the two-dimensional model.

**SUMMARY AND CONCLUSION**

A three-dimensional model for calculation of rearfoot and lower leg angles during running was developed. Two different relevant sequences of Euler angles between the foot and the lower leg were investigated. As the movement about the subtalar joint is small during running, the difference between the two three-dimensional alternatives proposed for calculation of movements between lower leg and foot is also small. In more extreme movements there would be a larger difference. To avoid confusion one of them should become standard. If the motion were calculated about the anatomically more relevant rotation axes, ankle joint and subtalar joint, therefore would become standard. If the motion were calculated about the anatomically more relevant rotation axes, ankle joint and subtalar joint, there would not be any source of confusion, but that would demand some knowledge about the rearfoot anatomy of every test subject.

The reliability of the measurement technique expressed as standard deviation of an important variable (maximal eversion angle) was investigated. The errors due to re-calibration of the measurement system including re-calibration of the segment coordinate systems of the test subject were well below the standard deviation among single steps. It is concluded that the described technique is useful for measuring and describing rearfoot and lower leg motion.

With the two-dimensional technique commonly used, angles are calculated about coordinate axes fixed outside the body, giving angle values without a simple relationship to anatomical movements. This is avoided by using the described three-dimensional method. The magnitude of this problem was investigated. Two-dimensional angles and variables were calculated and compared with corresponding three-dimensional parameters at different alignment angles between the foot and the camera axis. It was shown that most two-dimensional angular values measured from a posterior view were very sensitive to the alignment angle. The relative eversion angle, $\Delta\beta_{\text{pro}}$, was the variable that was least sensitive to the alignment angle.

Two-dimensional angular values measured from a lateral view are not sensitive to foot/camera alignment and provide a result well in line with the three-dimensional results.

It is concluded that the two-dimensional angles of running measured from posterior views are influenced by normal rotations about other axes and projection errors to a considerable extent. It would be difficult to compensate for these properties to get an anatomically more meaningful result. It remains to be shown which three-dimensional variables are correlated to injuries.
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REFERENCES


