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Repeatability of a model for measuring multi-segment foot kinematics in children

J. Stebbins *, M. Harrington, N. Thompson, A. Zavatsky, T. Theologis

Oxford Gait Laboratory, Nuffield Orthopaedic Centre, Windmill Road, Headington, Oxford OX3 7LD, UK

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Abstract

This study used a previously tested foot model and adapted it for use with children. A number of variations on this adapted model were implemented and tested for repeatability and accuracy on 15 healthy children on three occasions. These included redefinition of the long axes of the tibia and forefoot, assessment of the flexibility of the forefoot and evaluation of the variability of the wand marker on the heel for both static and dynamic trials. It was found that variations on the model produced only minimal changes in repeatability, the only significant change being elimination of the wand marker on the heel in the static trial, which reduced between-day variability of hindfoot motion in the transverse plane. However, some differences were evident in the mean values for all variations. Based on these results, the most accurate and appropriate version of the model is proposed, and average kinematic curves are presented based on the measurements from 14 healthy children.

Keywords: Foot; Kinematics; Children; Repeatability

1. Introduction

The measurement of three-dimensional lower body kinematics through the use of stereophotogrammetry has been well tested and validated over many years. Gait analysis now forms a major part of clinical decision making, particularly in the field of surgical treatment for children with cerebral palsy [1]. Conventional lower body models represent the pelvis, femur and lower leg as separate rigid bodies; however, the foot is routinely modelled only as a single vector, with no relative motion between or within its different segments. This provides inadequate information when determining treatment specific to the foot.

Measurement of foot kinematics is becoming increasingly common as motion analysis measuring systems become more accurate. Many research groups around the world are proposing multi-segment foot models, and it is important that the repeatability and clinical significance of these models be thoroughly investigated before they are routinely used to inform clinical decision-making. It is also necessary to standardise the analysis and reporting of results to allow comparisons between centres.

Most work to date has been carried out on healthy adult feet [2–11] and these studies are mainly limited to the stance phase of gait. A limited amount of work has also been conducted on pathological feet [7,12–14], again mainly in adults. No repeatability studies have been reported for measurements on children's feet. This population poses different challenges, the most significant being the small surface area of the foot and greater variability in gait [15]. There are many conditions that produce deformity in children's feet. For example, 90% of children with cerebral palsy develop some form of foot deformity resulting from abnormal forces being applied to the immature skeleton over periods of growth [16]. A valid and repeatable foot model for children is needed for understanding normal and pathological function, planning intervention and evaluating the outcome of treatment.

The purpose of the current study was to take our previously described multi-segment foot model validated for healthy adults [4] and adapt it for use in children over the entire gait cycle. Five variations of the adapted model were then tested to determine the most appropriate method for

* Corresponding author. Tel.: +44 1865 227609.
E-mail address: julie.stebbins@ndos.ox.ac.uk (J. Stebbins).

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measuring inter-segment motion within the foot for both healthy and pathological conditions, such as cerebral palsy.

2. Materials and method

Fifteen healthy children aged 6–14 years (average age 9.5 years, 5 male and 10 female) were tested at the Oxford Gait Laboratory on three separate occasions. Visits were spaced between 2 weeks and 6 months apart. Each child had reflective markers placed on their dominant foot [4] and a conventional marker set on the lower body [17]. A 12 camera VICON 612 system (Vicon Motion Systems Ltd., Oxford, UK) was used to collect 3D kinematics of one foot and both lower limbs for each subject at 100 Hz. A static standing trial was performed to define the segment axes before three markers were removed (Table 1) for the walking trials. Subjects were asked to walk at their usual speed along a 10-m walkway. Three representative strides from three separate trials were used for analysis from each session. These trials were identified visually by looking at all traces from the session (average 20 trials). A static trial was repeated at the end of the session as a check against any markers being displaced.

The Oxford Foot Model defined by Carson et al. [4] was modified for children and for application to deformities seen in cerebral palsy. The original model comprised a rigid tibial segment (tibia and fibula), a rigid hindfoot (calcaneus and talus), a forefoot (five metatarsals) and a hallux (proximal phalanx of the hallux). The tibia was redefined using the knee joint centre for compatibility with the lower body model. The hindfoot segment was altered to be independent of neighbouring segments, which is particularly important in cases where foot deformity is present. For the forefoot, the position of the proximal marker on the first metatarsal (P1MT) was changed to be medial to the extensor hallucis longus tendon, to provide greater consistency in placement; the forefoot anterior/posterior axis was adjusted accordingly. A single vector replaced the hallux segment. This gave the following foot segment definitions for what was considered to be the *default model*, with markers listed in Table 1 and shown in Fig. 1a and b.

Table 1
Names and positions of markers used in the foot model

Marker name	Position	Segment
KNE	Femoral condyle	Femur
TTUB	Tibial tuberosity	Tibia
HFIB	Head of fibular	Tibia
LMAL	Lateral malleolus	Tibia
MMAL	*Medial malleolus*	*Tibia*
SHN1	Anterior aspect of shin	Tibia
CAL1	Posterior distal aspect of heel	Hindfoot
CAL2	*Posterior medial aspect of heel*	*Hindfoot*
CPEG	Wand marker on posterior calcaneus aligned with transverse orientation	Hindfoot
LCAL	Lateral calcaneus	Hindfoot
STAL	Sustentaculum tali	Hindfoot
P1MT	Base of first metatarsal	Forefoot
P5MT	Base of fifth metatarsal	Forefoot
D1MT	*Head of first metatarsal*	*Forefoot*
D5MT	Head of fifth metatarsal	Forefoot
TOE	Between second and third metatarsal heads	Forefoot
HLX	Base of hallux	Hallux

Names in italics are used in the static trial only and are removed for dynamic trials.

2.1. Tibia

2.1.1. Anatomical definition

Composed of the tibia and fibula, and assumed to move as a single rigid body. Segment is based on the plane defined by the line from the knee joint centre to the ankle joint centre and the trans-malleolar axis. Longitudinal axis is from the ankle joint centre to the knee joint centre. Anterior axis is perpendicular to the plane defined by the longitudinal axis and the trans-malleolar axis. Transverse axis is mutually perpendicular.

2.1.2. Placement of markers

HFIB placed on the most lateral aspect of the head of the fibula. TTUB placed on the most anterior aspect of the tibial tuberosity. SHN1 placed anywhere along the anterior crest the tibia. LMAL placed on the most lateral aspect of the lateral malleolus. MMAL placed on the most medial aspect of the medial malleolus.

2.1.3. Technical frame

Axis 1 from HFIB to LMAL. Axis 2 perpendicular to axis 1 and the vector from SHN1 to the mid-point between LMAL and HFIB.

2.1.4. Static calibration

Vertical axis from mid-point between MMAL and LMAL to KJC (defined by the lower body model [17]). Anterior axis perpendicular to the plane defined by vertical axis and the vector from MMAL to LMAL. Transverse axis mutually perpendicular.

2.2. Hindfoot

2.2.1. Anatomical definition

The calcaneus defines the hindfoot. Motion at the talocrural and sub-talar joints are considered to contribute jointly to motion of the hindfoot relative to the tibia. Segment is based on orientation of the mid-sagittal plane of the calcaneus in standing posture (defined by the line which passes along posterior surface of the calcaneus, which is equidistant from both lateral and medial borders of this surface, and the point mid-way between the sustentaculum tali and the lateral border of the calcaneus). Anterior axis from the most posterior aspect of the calcaneal tuberosity, in the plane defined above and parallel to the plantar surface of the hindfoot. Lateral axis perpendicular to this plane.

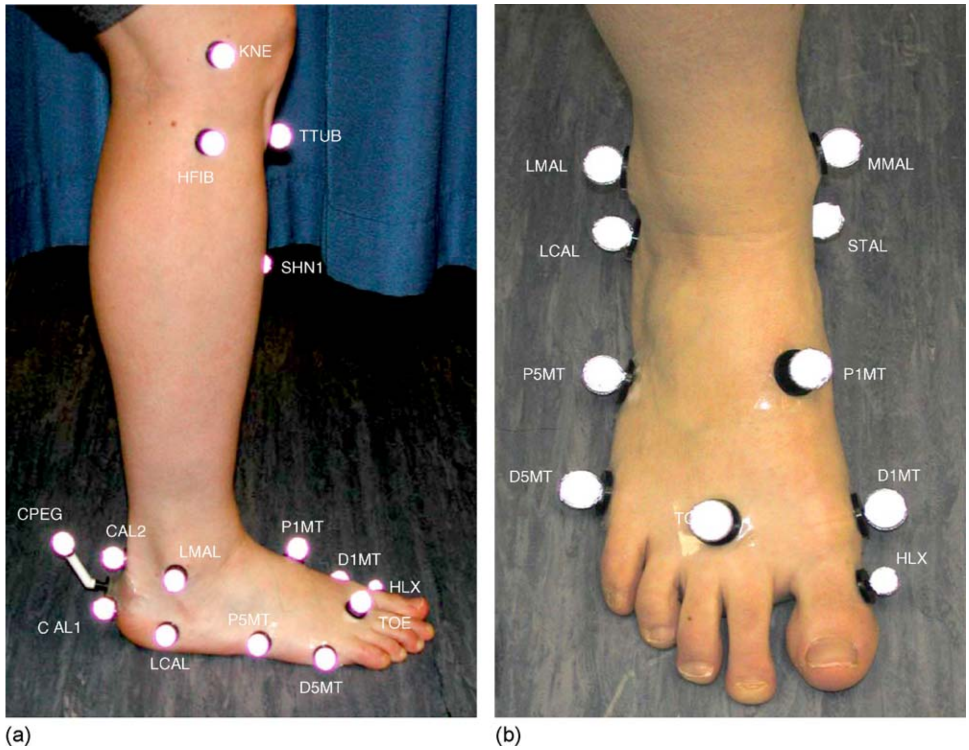

Fig. 1. (a) Markers on the lower leg and foot shown from a lateral view. (b) Markers on the foot and ankle shown from a dorsal view.

2.2.2. Placement of markers

CAL1 and CAL2 placed on posterior aspect of calcaneus such that they are on the distal and proximal ends of the mid-line in the sagittal plane, respectively. CPEG is a wand marker, the base of which is placed mid-way between CAL1 and CAL2, and the marker itself in the same line as CAL1 and CAL2, and in line with transverse orientation of hindfoot. STAL placed on the sustentaculum tali. LCAL placed on the lateral aspect of the calcaneus, at the same distance from the most posterior point as STAL.

2.2.3. Technical frame

Axis 1 from CAL1 to CPEG, axis 2 perpendicular to axis 1 and the vector from CAL 1 to LCAL.

2.2.4. Static calibration

Anterior axis is parallel to the floor and in the plane of CAL1, CAL2 and CPEG. Transverse axis is perpendicular to this plane. Vertical axis is mutually perpendicular.

2.3. Forefoot

2.3.1. Anatomical definition

Comprised of the five metatarsals, and assumed to move as a single rigid body. Segment is based on the plane defined by the centre of distal heads of the first and fifth metatarsals and the proximal head of the fifth metatarsal. Superior axis is perpendicular to this plane.

Longitudinal axis is the projection of the line from the mid-point of the proximal heads of the first and fifth metatarsals to the mid-point of the distal heads of the second and third metatarsals into this plane.

2.3.2. Placement of markers

D1MT and D5MT placed medially and laterally on the foot such that their centres lie along the line through the distal heads of the first and fifth metatarsal heads. P5MT placed laterally over the proximal head of the fifth metatarsal in the plane containing this and markers D1MT and D5MT. TOE placed of the mid-point of the distal heads of the second and third metatarsals. P1MT placed on proximal head of first metatarsal, just medial to the extensor hallucis longus tendon (palpated by asking the subject to dorsiflex the hallux).

2.3.3. Technical frame

Axis 1 from P1MT to D5MT. Axis 2 perpendicular to this and the vector from TOE to P5MT.

2.3.4. Static calibration

Vertical axis perpendicular to plane containing D1MT, D5MT and P5MT. Longitudinal axis is the projection onto this plane of the line from the mid-point of P1MT and P5MT to TOE. Transverse axis is mutually perpendicular.

2.4. Hallux

2.4.1. Anatomical definition

Comprised of the proximal phalanx of the hallux. Based on a longitudinal line along the proximal phalanx.

2.4.2. Placement of markers

HLX placed on the medial side of the proximal phalanx of the hallux, mid-way between the superior and inferior surface.

2.4.3. Static calibration

Medio-lateral axis aligned with lateral axis of forefoot. Anterior axis from D1MT (at height of HLX) to HLX. Angles of rotation for each segment were calculated according to the joint co-ordinate system of Grood and Suntay [18].

For the hindfoot relative to the tibia (HINDFOOT/TIBIA)

- Plantar/dorsiflexion: about the transverse axis of the tibia.
- Inversion/eversion: about the longitudinal axis of the hindfoot.
- Internal/external rotation: about an axis perpendicular to the previous two axes.

For the forefoot relative to the hindfoot (FOREFOOT/HINDFOOT)

- Plantar/dorsiflexion: about the transverse axis of the hindfoot.
- Supination/pronation: about the longitudinal axis of the forefoot.
- Abduction/adduction: about an axis perpendicular to the previous two axes.

For the hallux relative to the forefoot (HALLUX/FOREFOOT)

- Plantar/dorsiflexion: about the transverse axis of the forefoot.

To compare results with conventional lower body models that represent the foot as a single vector, rotations of the *forefoot relative to the tibia (FOREFOOT/TIBIA)* were also calculated:

- Plantar/dorsiflexion: about the transverse axis of the tibia.
- Supination/pronation: about the longitudinal axis of the forefoot.
- Abduction/adduction: about the axis perpendicular to the previous two axes.

This model was implemented using BodyBuilder software (Vicon Motion Systems Ltd., Oxford, UK). The four markers on each segment (except the hallux) give redundancy in case of marker loss. This was achieved through use of a REPLACE function within BodyBuilder, which estimates the position of a missing marker on a rigid segment on the basis of the remaining three visible markers. This information was also exploited to calculate inter-segment angles in different ways. Five variations of the default model were tested for anatomical feasibility and repeatability:

1. Using a scaled virtual point between D1MT and D5MT to define the distal point of the longitudinal axis of the foot (DistFF). The point was chosen as the scaled distance that produced the same angle of forefoot abduction as using the TOE marker. It was thought this would prove more reliable than placing the TOE marker on the foot. This altered the *static calibration* to be the following: vertical axis perpendicular to plane containing D1MT, D5MT and P5MT. Longitudinal axis is the projection onto this plane of the line from the mid-point of P1MT and P5MT to the point 1/1.9 between D1MT and D5MT. Transverse axis mutually perpendicular.
2. Tracking the forefoot segment with markers on the lateral part only—anatomically defined as being comprised of metatarsals two to four (eliminating the use of P1MT for dynamic trials). This resulted in the following altered *technical frame:* axis 1 from D5MT to P5MT. Axis 2 perpendicular to this and the vector from TOE to D5MT. To quantify the sensitivity of the measurements to movement of P1MT relative to the forefoot (i.e. flexibility of the forefoot segment in the default model), P1MT was moved up and down by 5 mm within the forefoot co-ordinate system of one representative subject, using BodyBuilder software (Vicon Motion Systems Ltd., Oxford, UK). The resulting changes on forefoot pronation and supination were calculated. A value for forefoot "arch height" of the foot was also calculated. This was defined as the absolute distance between the marker on the dorsal surface of the base of the first metatarsal (P1MT), and the same point projected on the plantar surface of the forefoot as defined by the model in the static trial. This was then normalised to foot length by dividing by the distance from CAL1 to TOE, which gave the arch height as a percentage of foot length. The maximum and range of this distance were recorded.
3. Tracking the hindfoot segment in walking trials without the use of the wand marker on the rear of the calcaneus (CPEG). This was performed since it appeared that the wand marker was prone to being knocked, and therefore, a potential source of variability. The following altered *technical frame* was used: axis 1 from CAL1 to the mid-point between LCAL and STAL. Axis 2 perpendicular to axis 1 and the vector from CAL1 to LCAL.
4. Using markers on the tibia itself to define the longitudinal axis of the tibia (as in the original version of the model [4]), rather than the knee joint centre from the conventional lower body model. This was also believed to potentially improve repeatability, since the assumption

of a fixed knee joint centre may be a source of variability. This resulted in the following alteration to the *static calibration*: vertical axis from mid-point between MMAL and LMAL to projection of TTUB into the plane defined by MMAL, LMAL and HFIB. Anterior axis perpendicular to the plane defined by vertical axis and the line from MMAL to LMAL. Transverse axis mutually perpendicular.

5. Static calibration of hindfoot and dynamic tracking of hindfoot without reference to CPEG (except to use in REPLACE command). It was thought this would prove to be more accurate and repeatable than visually lining up the CPEG with the orientation of the hallux. This altered the *static calibration* to be the following: anterior axis is parallel to the floor and in the plane of CAL1, CAL2 and the mid-point between STAL and LCAL. Transverse axis is perpendicular to this plane. Vertical axis is mutually perpendicular.

The *technical frame* was the same as in point 3 above.

Parameters that were analysed included clinically significant maximums, minimums and ranges of inter-segment angles. These were determined after an initial visual analysis of the data, as well as consideration of routine clinical practice. Data from three trials for each visit were averaged and then one-way ANOVA tables were used to determine within subject standard deviations to give a measure of repeatability. Each variation of the model was compared to the default model. Changes in within-subject standard deviations greater than 1° were considered significant. Mean values for each variable were also calculated and differences were assessed with paired *t*-tests ($p < 0.01$).

3. Results

Measured variables from foot kinematics for each subject were compared between days. For the default model, the patterns of movement were found to be consistent but some offsets between days were observed (Fig. 2). Within subject standard deviations were lowest in the sagittal plane (between 2° and 4°). The highest variability was in the transverse plane at the hindfoot and forefoot; with standard deviations of 8° and 7°, respectively.

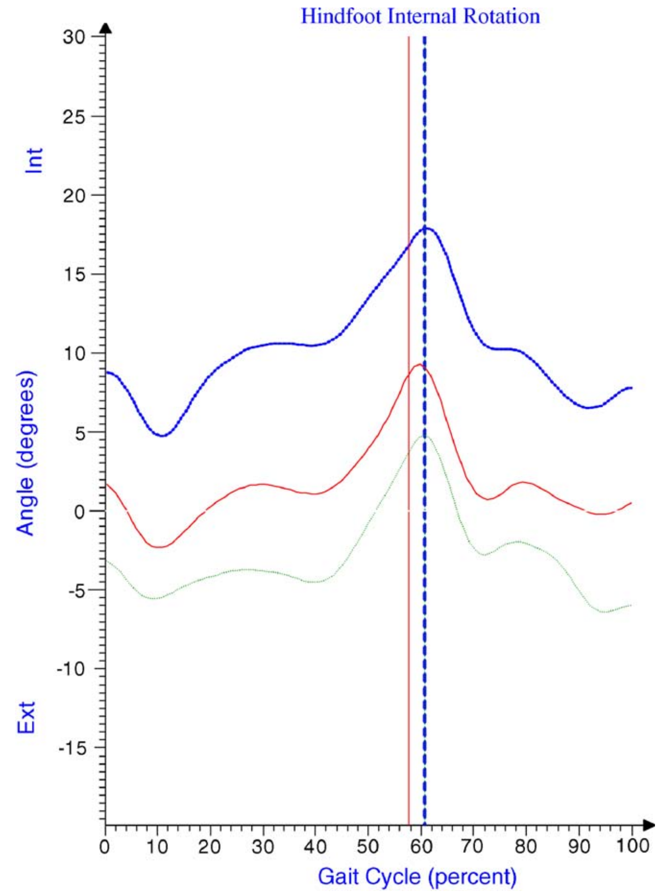

Fig. 2. Example of inter-day variability in foot motion. Patterns consistent but offset apparent. Solid line from visit 1, dashed line from visit 2, dotted line from visit 3. Vertical lines divide stance and swing phase.

Table 2
Comparison of means and within subject standard deviations for the default model and variations (1) and (2), which affect the position of the forefoot relative to the hindfoot and tibia

	Default		DistFF point (1)		Lateral markers only (2)	
	Mean	S.D.	Mean	S.D.	Mean	S.D.
Maximum forefoot dorsiflexion	9.8	3.4	9.8	3.3	10.1	3.3
Maximum forefoot supination	6.5	5.3	6.4	5.9	6.5	5.4
Minimum forefoot abduction	5.0	7.4	4.1	7.1	0.6[a]	7.1
Range forefoot dorsiflexion	20.8	2.7	20.8	2.7	19.2[a]	2.5
Range forefoot supination	8.8	1.6	8.6	1.6	7.8[a]	2.0
Range forefoot adduction	9.9	2.4	9.9	2.6	8.8[a]	2.3
Maximum FF/Tib dorsiflexion	19.9	3.7	19.9	3.5	NA	NA
Maximum FF/Tib supination	11.8	4.2	12.8	3.4	NA	NA
Maximum FF/Tib adduction	16.8	8.4	19.6	7.8	NA	NA
Range FF/Tib dorsiflexion	44.0	4.5	44.1	4.5	NA	NA
Range FF/Tib supination	15.2	3.1	16.4	3.3	NA	NA
Range FF/Tib adduction	17.3	3.7	18.4	3.4	NA	NA

[a] Indicates significant difference in the mean value.

Table 3
Comparison of means and within subject standard deviations for the default model and variations (3–5) which affect the hindfoot

	Default		No CPEG (3)		Tibia markers (4)		No CPEG static (5)	
	Mean	S.D.	Mean	S.D.	Mean	S.D.	Mean	S.D.
Maximum knee flexion	59.2	2.5	NA	NA	64.5*	3.1	NA	NA
Maximum hindfoot dorsiflexion	11.2	3.0	10.9	2.9	12.2	3.5	10.7	2.8
Maximum hindfoot inversion	9.0	5.2	8.8	5.4	9.3	5.6	7.6	4.6
Maximum hindfoot rotation	13.8	8.4	15.6	8.3	15.3	8.1	13.9	6.6[b]
Range knee flexion	56.9	3.0	NA	NA	61.2[a]	3.2	NA	NA
Range hindfoot dorsiflexion	24.2	2.7	23.7[a]	2.8	20.4[a]	2.6	23.8	2.8
Range hindfoot inversion	10.8	2.2	11.5	2.1	10.2	1.9	11.4	2.0
Range hindfoot rotation	12.3	2.7	12.4	2.0	11.1	2.0	12.1	2.1

[a] Indicates significant difference in mean value.
[b] Indicates significant difference in repeatability.

All five variations of the model were compared for repeatability to the default model. Changes in definitions of the model produced only minimal changes in repeatability. The only variation producing change in the repeatability was variation 5, which reduced variation of the hindfoot in the transverse plane between days from 8° to 6° (see Tables 2 and 3).

3.1. Variation 1

Using a scaled point 53% of the distance between the heads of the first and fifth metatarsals produced no significant change in the measured angle of maximum forefoot adduction relative to the hindfoot, when compared to using the TOE marker (Table 2). It was, therefore, accepted that this scaled distance produced similar results to the TOE marker and was used for further comparison. No difference was noted in repeatability in using this point instead of the TOE marker.

3.2. Variation 2

Using markers on the lateral part of the forefoot only to track the forefoot had the effect of making the forefoot appear less adducted in swing than if markers on the entire forefoot were used (mean change from 5° to 1° adduction, $p < 0.01$) (Fig. 3). It also reduced the range of forefoot motion (dorsi/plantar flexion: 21–19°; abduction/adduction: 9–8°; and pronation/supination: 10–9°, $p < 0.01$) (Table 2). The calculated forefoot ‘‘arch height’’ had a mean value of

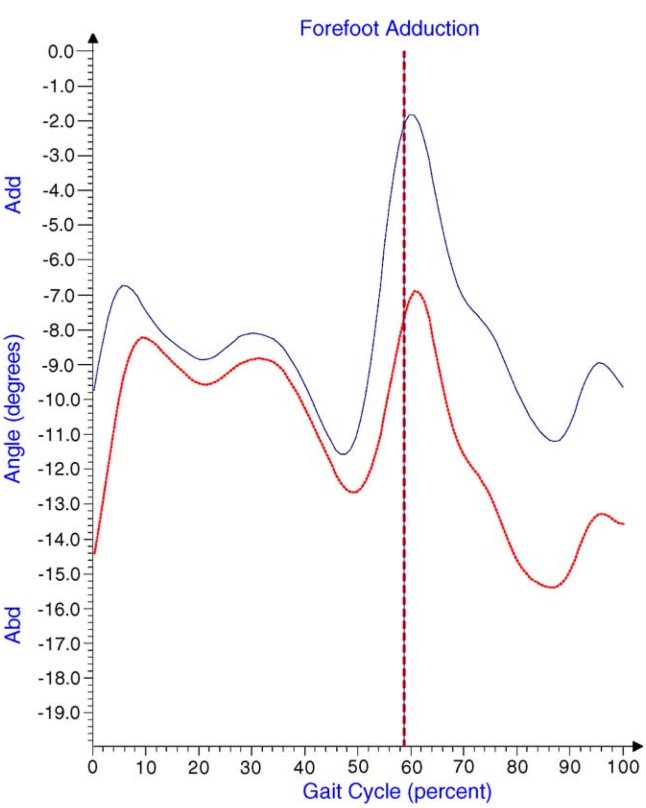

Fig. 3. Change forefoot abduction when using lateral markers only to track the forefoot (dashed line). Solid line represents tracking all four markers on the forefoot (default model).

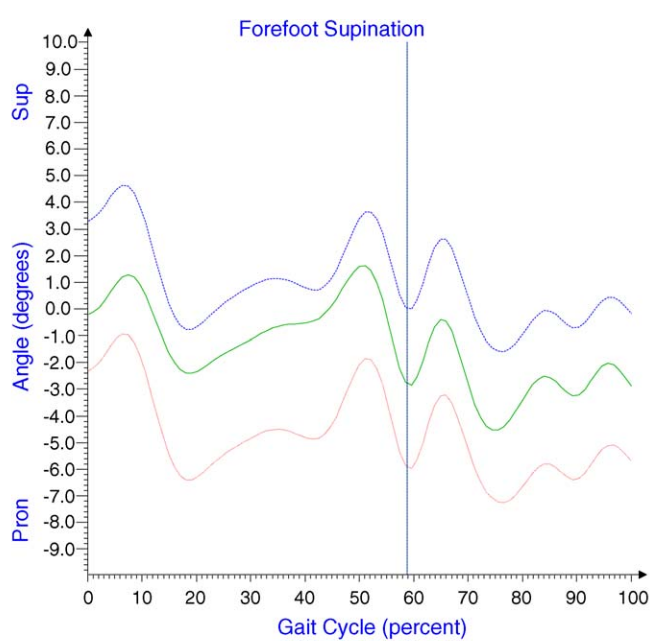

Fig. 4. Change in forefoot angles as a result of shifting P1MT up (dashed line) by 5 mm and down (dotted line) by 5 mm. Solid line is neutral position of P1MT as in default model. Vertical dashed line divides stance and swing phase. Trace is from one representative subject.

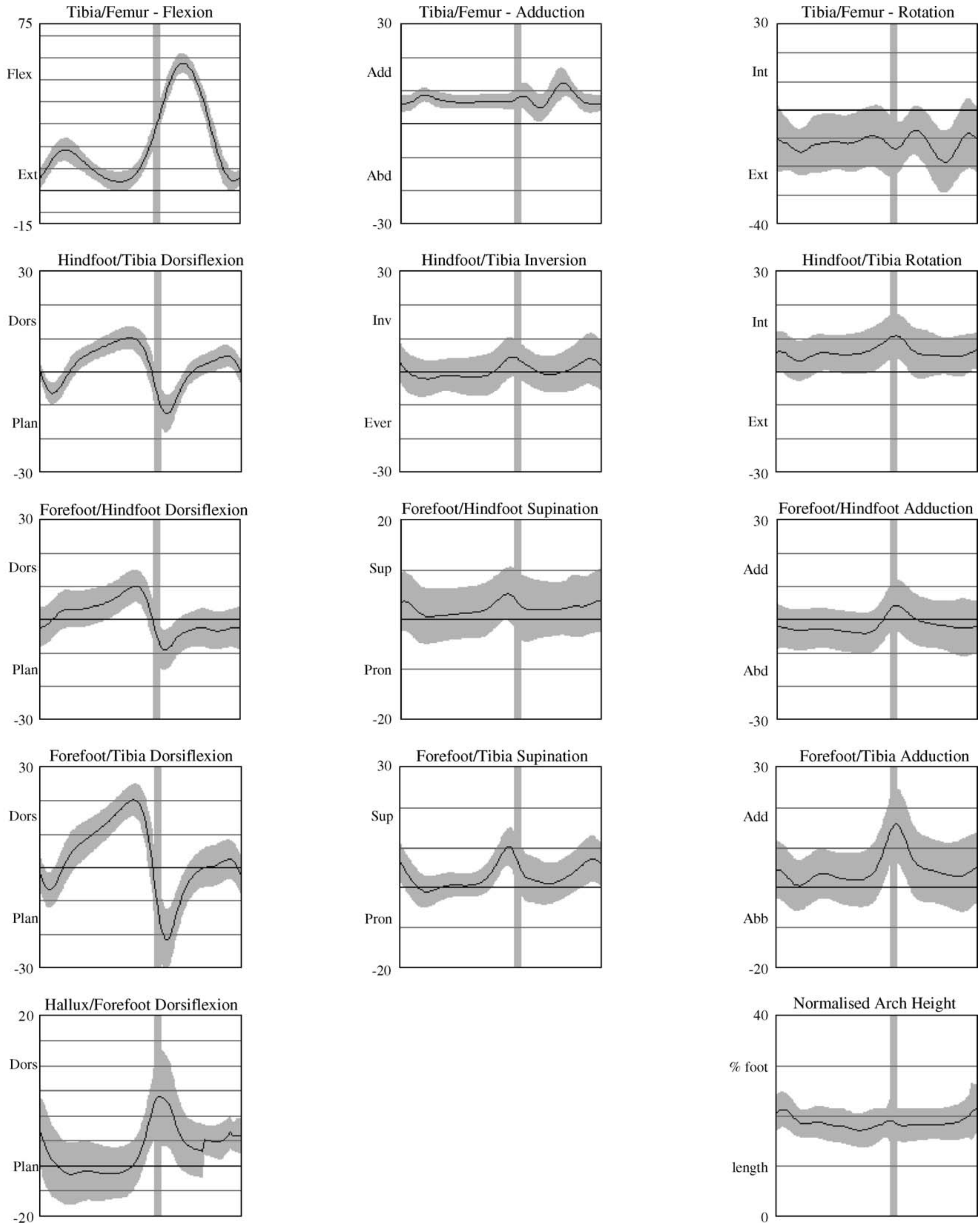

Fig. 5. Inter-segment foot angles from 14 healthy children of final version of Oxford foot model, normalised to 100% of the gait cycle. Shade band shows mean ± 1 standard deviation across all subjects.

23% (S.D. 4%). This changed within the gait cycle by 7% (S.D. 2%). Sensitivity analysis on the position of the proximal marker on the first metatarsal revealed that a movement of 10 mm in the vertical direction produced a maximum change of 9° in supination of the forefoot (Fig. 4).

3.3. Variation 3

When the wand marker on the rear of the calcaneus was eliminated from use in calculating hindfoot angles in dynamic trials, repeatability was unchanged. However, it tended to reduce the range of hindfoot dorsiflexion (from 24.2° to 23.7°, $p < 0.01$) (Table 3).

3.4. Variation 4

Using anatomical markers on the tibia itself, rather than the knee joint centre, produced a significant increase in the mean maximum knee flexion angle (from 59° to 65°, $p < 0.01$) and range of knee flexion (from 57° to 61°, $p < 0.01$). It also reduced the range of measured hindfoot dorsiflexion (from 24° to 20°, $p < 0.01$) (Table 2).

3.5. Variation 5

Elimination of the CPEG wand marker during static calibration was the only variation to significantly improve repeatability, which reduced the standard deviation for hindfoot rotation from 8° to 6°. There no significant changes in mean values.

Table 2 shows a summary of the comparison of within subject standard deviations for each of the methods described above, as well as comparisons of mean values. Fig. 5 shows the average foot kinematics from 14 healthy children based on the default version of the model. One subject was excluded from this average due to obvious toe walking.

4. Discussion

A comprehensive study has been carried out on the repeatability of foot kinematics in healthy children and this has provided objective data on which to base a final version of the model.

For the default model, good consistency between patterns of foot motion was observed but some offsets in the curves between days were noted. This was quantified by comparing intra-subject standard deviations in peak and range values for the inter-segment angles where the peak values between adjacent segments showed greater variability (2–8°) than the ranges (2–3°). The sagittal plane (dorsi/plantarflexion) was found to be the most repeatable and the highest variability was found in the transverse plane (internal/external rotation of the hindfoot and forefoot ab/adduction). This variability is generally greater than previously reported values based tests in adult subjects, as might be expected due to the smaller surface area of children's feet and the greater variability in their gait [14]. The trend in variability between planes (between 1° and 5°) is consistent with the results previously found when validating the Oxford Foot Model on two healthy adults [4]. Other investigators who have studied adult foot kinematics have found similar trends. Siegel et al. [10] reported similar standard deviations, with the sagittal plane having the least variability, followed by the coronal plane and the transverse plane with the most variability. Moseley et al. [11] found 1–2° standard deviations, again with the greatest repeatability appearing in the sagittal plane and the least in the transverse plane. Kepple et al. [9] tested five healthy adults and found intra-subject standard deviations ranging up to 1.2°. They found the most variability in the transverse plane, and the least in the coronal plane. Leardini et al. [8] reported acceptable levels of repeatability in the sagittal and coronal planes, but not in the transverse, for all inter-segment pairings. They further stated that clinical conclusions on altered joint function in the transverse plane should, therefore, be handled more carefully. There have been some studies that found the most variability in the coronal plane [6,7]. The three studies of pathological feet [7,12,13] all found greater variability in the clinical cases when compared to the healthy controls.

It is difficult to compare directly results with other studies since there are differences between the definitions of axes and testing protocols. Most models [8,11] reference their dynamic angles to a standard neutral position, where all joint angles are defined to be zero. This would artificially reduce variation between sessions, as shifting the reference position each time reduces any error due to marker placement. The Oxford Foot Model does not reference to a measured neutral posture, but instead allows non-zero joint angle in the static reference position. This would account for some of the increased variability, but allows for measurement of foot deformity where achieving "neutral position" in a static standing posture is not always possible. Repeatability testing was also carried out on the same day by several researchers [2,5–8,10], whereas the study described here was repeated on three separate days, which would also be likely to reduce repeatability. Furthermore, the method of quantifying repeatability differs between authors. In the current study, the variability in peak and range values are reported since these were considered to have clinical significance, whereas most previous studies report averaged values across a gait cycle or stance phase [8,19].

The resulting differences in the repeatability from changes to the model tested within the current study were generally small. When the mid-point between the head of the first and fifth metatarsal was used to define the longitudinal axis of the forefoot (instead of an actual "TOE" marker), repeatability was unchanged in all measured variables. Virtual markers have been reported to have greater reliability than physical markers [20], and it was thought that identification of points on either side of the foot, rather than one on the top of the foot would prove to be more accurate. This may still be the case in feet with significant deformity, where the metatarsals are no longer parallel, and identification of a point between the second and third metatarsal heads may not represent the long axis of the forefoot. It may, therefore, be useful to maintain this as an option to calculate forefoot angles (since there is no significant difference in mean angles produced by this method and by using the TOE marker). Leardini et al.'s model [8] suggests a maximum forefoot adduction of around 10° as do Rattanaprasert et al. [7]. Hunt et al. [6] suggest a maximum of 2° in their model of forefoot motion, which is similar to Kidder et al. [5], Wu et al. [13] and MacWilliams et al. [2]. Although these studies concerned adult feet, the results are similar to those found in the current study.

It was decided to attempt to track the forefoot with markers on the lateral part only to assess the relative movement of the P1MT marker and determine the validity of assuming this segment to be rigid. It was found that tracking with lateral markers only did not significantly improve repeatability. However, the measurement of "arch height" relative to the lateral forefoot was found to vary by an average of 7% of foot length within the gait cycle across all subjects. With the default definition of the forefoot that considers the five metatarsals as a rigid body, this "arch height" measure would be constant. The variability in this measure is, therefore, due to flexibility between the medial and lateral forefoot. On this basis, a sensitivity analysis was conducted on one representative subject. Since this subject had a foot length of 180 mm, a change in arch height of 7% foot length would be equal to 12.6 mm. It was, therefore, concluded that P1MT was likely to move by at least this amount during the gait cycle. The effect on pronation and supination of the forefoot was assessed by shifting the position of P1MT up and down by 5 mm. It was discovered that the maximum effect of moving the marker up by 5 mm produced an increase in supination of 5°, while moving the marker down by 5 mm produced a reduction in supination of 4°, making a range of 9°. This suggests that the error induced in forefoot supination, by assuming the rigidity of the segment, may be as high as 9°, when the variation in forefoot "arch height" is around 7% of foot length. As a result of this analysis, it was decided to routinely record this measure of "arch height" and to assume that the value of range in "arch height" across the gait cycle (% foot length) roughly corresponds to the error in supination of the forefoot (°) when measuring it as one rigid segment.

The anterior axis of the hindfoot was assumed to be parallel to the floor for this study, since it involved only healthy children, all of whom were able to stand with feet flat on the floor. Where this is not achievable, a rotational offset in the sagittal plane may be input (on the basis of X-ray or additional marker position).

It was initially thought that using a wand marker on the rear of the calcaneus was the source of variability in hindfoot angles, since it was likely to be knocked out of place. When the repeatability of measuring hindfoot motion without this wand marker was examined, it was found that there was no significant change in repeatability when this marker was eliminated. When not using the wand, a lower range of measured dorsiflexion of the hindfoot was apparent. However this value was very small (less than a degree) and not considered clinically relevant. It was decided to maintain the use of the wand marker, but to keep the option of tracking without it, in cases of obvious marker loss or knock, which can be assessed by comparing results from pre and post assessment static trials.

Using physical markers on the tibia itself, rather than the calculated knee joint centre from the conventional lower body model [17], was expected to improve repeatability of measurements, but no significant change was noted. However, it significantly altered the absolute values of three of the eight kinematic variables. Primarily, it changed the apparent amount of knee flexion, and also affected range of hindfoot dorsiflexion. Maximum knee flexion was increased from 59° to 65° when markers on the tibia were used. The change in knee flexion angle when the knee joint centre is not used may reflect translational error in the tibia and this effect may warrant further investigation. However, for consistency with conventional kinematics, it was decided to continue to use the knee joint centre to align the long axis of the tibia, to be compatible with conventional lower body kinematic models.

When the CPEG marker was eliminated from the static calibration, an improvement of 2° in the within subject standard deviation was noted. It was thought that variability in aligning the CPEG marker with the transverse orientation of the hindfoot was the source of some variability in measuring of hindfoot rotation. The hypothetic line from CAL1 through the middle of the calcaneus, is used to visually align the CPEG marker, so it was thought that directly using the line from CAL1 to the mid-point between STAL and LCAL (placed on either side of the calcaneus) would be a more repeatable estimate of the anterior axis of the hindfoot. Whilst it is possible that this may reduce the sensitivity of the model to identifying deviation in the transverse plane, it is believed this is outweighed by the improvement in repeatability gained by using the mid-point between STAL and LCAL instead of the CPEG marker. This assumption will require further validation on feet with deformity of the hindfoot in the transverse plane.

The terminology for this model was adopted on the basis of clinical convenience and clarity. It is recognised that axes of rotation defined by the model do not always exactly replicate true anatomical axes. For example, Leardini et al. [21] confirmed in vitro that true inversion/eversion takes place at the sub-talar or talocalcaneal joint about an axis whose position is load-dependant. The present foot model combines the calcaneus and talus into a single segment (the hindfoot) and hence is unable to separate true inversion from overall relative motion between the tibia and hindfoot. However, the term ‘‘inversion/eversion’’ is meaningful to clinicians using this model, but it needs to be taken in the context of the definition described.

Taking into account repeatability and compatibility issues with the existing lower body model, it was decided to continue using the physical ‘‘TOE’’ marker to calculate the long axis of the forefoot, to eliminate use of the CPEG wand marker on the calcaneus, to use the conventional knee joint centre to calculate the long axis of the tibia, and to measure forefoot ‘‘arch height’’ relative to the plantar surface defined by lateral markers on the forefoot to allow estimation of error produced in forefoot supination as a result of rigid body assumptions. Mean angles from the 14 healthy children of this version of the model are shown in Fig. 5.

An awareness of the variability in measurement of inter-segment foot motion in children is vital for correct interpretation of results and should not be ignored when planning treatment and assessing outcomes. While a number of different variations of the model were assessed to achieve the optimal model for measuring foot motion, up to 7° variability was still apparent in the transverse plane. It was recognised that this may be in part due to inherent variability in children's gait. However, a significant factor is the consistency of marker placement between days on small feet. Therefore, clear protocols and practice in marker placement are crucial and improvements to fixation of the reflective markers should be considered.

This study provided a validated multi-segment foot model for use with children, which previously was unavailable. The model proposed here produced results consistent with previous studies of foot kinematics in adults [2,4–7] and expected foot motion during normal gait [22]. This validation allows clinical implementation of the model, with an understanding of its reliability. The results in kinematic patterns were found to be more consistent than the absolute values. Absolute measurements in the transverse plane were found to be the least consistent, but repeatability was improved when the wand marker was eliminated from use on the hindfoot. The difference between measuring angles in slightly different ways gave only negligible differences in results, allowing for some flexibility in implementation in the presence of severe deformity.

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References

[1] Norlin R, Tkaczuk H. One-session surgery for correction of lower extremity deformities in children with cerebral palsy. J Pediatr Orthop 1985;5:208–11.
[2] MacWilliams BA, Cowley M, Nicholson DE. Foot kinematics and kinetics during adolescent gait. Gait Posture 2003;17:214–24.
[3] Woodburn J, Turner DE, Helliwell PS, Barker S. A preliminary study determining the feasibility of electromagnetic tracking for kinematics at the ankle joint complex. Rheumatology (Oxford) 1999;38:1260–8.
[4] Carson MC, Harrington ME, Thompson N, O'Connor JJ, Theologis TN. Kinematic analysis of a multi-segment foot model for research and clinical applications: a repeatability analysis. J Biomech 2001; 34:1299–307.
[5] Kidder SM, Abuzzahab Jr FS, Harris GF, Johnson JE. A system for the analysis of foot and ankle kinematics during gait. IEEE Trans Rehabil Eng 1996;4:25–32.
[6] Hunt AE, Smith RM, Torode M, Keenan AM. Inter-segment foot motion and ground reaction forces over the stance phase of walking. Clin Biomech 2001;16:592–600.
[7] Rattanaprasert U, Smith R, Sullivan M, Gilleard W. Three-dimensional kinematics of the forefoot, rearfoot, and leg without the function of tibialis posterior in comparison with normals during stance phase of walking. Clin Biomech 1999;14:14–23.
[8] Leardini A, Benedetti MG, Catani F, Simoncini L, Giannini S. An anatomically based protocol for the description of foot segment kinematics during gait. Clin Biomech 1999;14:528–36.
[9] Kepple TM, Stanhope SJ, Lohmann KN, Roman NL. A video-based technique for measuring ankle-subtalar motion during stance. J Biomed Eng 1990;12:273–80.
[10] Siegel KL, Kepple TM, O'Connell PG, Gerber LH, Stanhope SJ. A technique to evaluate foot function during the stance phase of gait. Foot Ankle Int 1995;16:764–70.
[11] Moseley L, Smith R, Hunt A, Gant R. Three-dimensional kinematics of the rearfoot during the stance phase of walking in normal young adult males. Clin Biomech 1996;11:39–45.
[12] Woodburn J, Helliwell PS, Barker S. Three-dimensional kinematics at the ankle joint complex in rheumatoid arthritis patients with painful valgus deformity of the rearfoot. Rheumatology 2002;41: 1406–1412.
[13] Wu WL, Su FC, Cheng YM, Huang PJ, Chou YL, Chou CK. Gait analysis after ankle arthrodesis. Gait Posture 2000;11:54–61.
[14] Theologis TN, Harrington ME, Thompson N, Benson MK. Dynamic foot movement in children treated for congenital talipes equinovarus. J Bone Joint Surg Br 2003;85:572–7.
[15] Stolze H, Kuhtz-Buschbeck JP, Mondwurf C, Johnk K, Friege L. Retest reliability of spatiotemporal gait parameters in children and adults. Gait Posture 1998;7:125–30.
[16] O'Connell PA, D'Souza L, Dudeney S, Stephens M. Foot deformities in children with cerebral palsy. J Pediatr Orthop 1998;18:743–7.
[17] Davis RB, Deluca PA. Clinical gait analysis. In: Harris GF., Smith P, editors. Human motion analysis. NJ, USA: IEEE Press; 1996. p. 17–42.
[18] Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. J Biomech Eng 1983;105:136–44.
[19] Liu W, Siegler S, Hillstrom HJ, Whitney K. Three-dimensional, six-degrees-of-freedom kinematics of the human hindfoot during stance phase of level walking. Hum Move Sci 1997;16:283–98.
[20] Soutas-Little RW. The use of ‘‘virtual markers’’ in human movement analysis. Gait Posture 1996;4:176–7.
[21] Leardini A, Stagni R, O'Connor JJ. Mobility of the subtalar joint in the intact ankle complex. J Biomech 2001;34:805–9.
[22] Perry J. Gait analysis: normal and pathological function. NJ: Slack International, 1992.