Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine
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Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine 2009 223: 981
DOI: 10.1243/09544119JEIM648

Comparing different data collection and analysis techniques for quantifying healthy knee joint function during stair ascent and descent

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The manuscript was received on 29 April 2009 and was accepted after revision for publication on 22 June 2009.

DOI: 10.1243/09544119JEIM648

Abstract: There is currently no standard data collection or analysis method for the assessment of stair gait using motion analysis. This makes the comparison of results from different studies difficult. It is important to gain an appreciation of the discrepancies in kinematic and kinetic information generated by employing different computational approaches, as these differences may be critical in cases where methodologies were to change over a long-term study. This study explores the effect of using different methodologies for the assessment of non-pathological knee function of ten subjects during stair ascent and descent. Two methods of computing knee kinematics were compared: (a) using in-house software and a pointer method of anatomical calibration and (b) using commercial software, Visual3D (C-motion, Inc.) and skin-mounted markers. Significant differences were found between the two methods when calculating a frontal plane range of motion ($p < 0.05$). Three methods of computing knee moments were compared. Knee moments computed using the inverse dynamic analysis (IDA) approach of Visual3D (C-motion, Inc.) were significantly different ($p < 0.05$) to those calculated using in-house IDA software that ignores the foot and ankle and to those computed using a vector cross-product approach. This study highlights the implications of comparing data generated from different collection and analysis methods.

Keywords: stair gait, knee kinematics, knee joint moments, motion analysis, biomechanics

1 INTRODUCTION

To identify changes in knee function associated with pathology and treatment it is important to assess the joint during a number of daily activities. Stair gait is commonly used as an assessment activity due to the large moments, forces, and ranges of motion at the knee joint required by the activity [**1**, **2**].

Motion capture synchronized with ground reaction force (GRF) measurements can quantify the kinematics and kinetics involved in daily activities. However, it is only useful as a clinical tool if accurate and practical assessments can be made using valid calculations and if similar outputs can be compared across a range of studies.

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There is currently no standard method of data collection or description of moments, making it difficult to compare results from different studies. Methodologies used by selected studies are summarized in Table 1. The range of methods includes using the vector cross-product approach of Andriacchi *et al.* [**18**], the calibrated anatomical system technique (CAST) [**8**], and the inverse dynamic analysis (IDA) of a linked segment model using commercial software.

During data collection using motion analysis, static calibrations are routinely performed prior to the measurement of dynamic movements to determine three landmarks per segment. These define segmental anatomical axes for the femur and tibia and their relationship with the thigh and shank external technical axes respectively. All subsequent articulations at the knee are measured using the technical axes. In previous studies, two methods

Table 1 Description of methodologies used to compute knee joint moments in selected studies

	Moment calculation
Selfe *et al.*, 2008 [**3**]	Standard inverse dynamic analysis (IDA) using Visual3D (C-motion, Inc.). Moments defined relative to the shank coordinate system
Protopapadaki *et al.*, 2007 [**4**]	Link segment method using Vicon Polygon software. Expressed as external moments
Thambyah *et al.*, 2004 [**5**]	IDA using VICON Clinical Manager Software (Oxford Metrics Limited)
Catani *et al.*, 2003 [**6**]	Moments determined according to the calibrated anatomical system technique (CAST) [**7**, **8**] and resolved into the joint coordinate system (JCS) [**9**]
Nadeau *et al.*, 2003 [**10**]	IDA [**11**] performed with Kingait3 software (Mishac Kinetics)
Costigan *et al.*, 2002 [**12**]	IDA ignoring the movement of the ankle joint. The shank's mass includes the mass of the foot and shoe. The shank's mass moment of inertia is modified using the principal axis theorem
Kaufman *et al.*, 2001 [**1**]	IDA using Orthotrack 4.0 (Motion Analysis Corp.)
Kowalk *et al.*, 1996 [**13**]	IDA using Gait lab software [**14**]. Moments defined in reference to the JCS
Yu *et al.*, 1996 [**15**]	IDA using Orthotrak II (Motion Analysis Corp.). Moments expressed in the tibia reference frame
McFadyen *et al.*, 1988 [**16**]	Uses 'BIOMECH' package [**17**] and calculations from reference [**11**]
Andriacchi *et al.*, 1980 and 1982 [**18**, **19**]	Cross-product of a vector defining the position of the joint centre and of the vector defining the GRF. Moments are resolved into the JCS

of anatomical calibration have been presented: (a) palpating bony landmarks and using a marked pointer to identify and record the bony prominences more accurately [**20**]; (b) the traditional method of placing markers on bony landmarks using a standard marker set. Pointer calibration data are generally reliant on the development of in-house software and are now also available in commercial software. Skin marker-based calibration [**6–8**] data are generally processed using commercial software such as Visual3D (C-motion, Inc.), with a linked model.

Knee joint moments are an important measure for stair gait and give an indication of how the muscles are functioning to control and stabilize the knee joint during the activity. Joint moments can be computed using different mathematical methods and are expressed as internal or external. This leads to confusion when attempting to validate a new set of measurements and prevents direct comparisons between studies.

This study used motion analysis and a previously reported staircase to compare tibiofemoral kinematics resulting from the two approaches to anatomical calibration. It also compared joint moments calculated using a vector cross-product approach and two inverse dynamics methods: one that ignores the foot and ankle effects and another that involves full inverse dynamics. Thus the current study objectives were to (a) compare two methods of computing knee joint kinematics and (b) compare three approaches of computing moments acting about the knee joint.

2 METHODS

2.1 Data collection

Motion analysis was performed for ten non-pathological (NP) subjects (i.e. with no known history of lower limb pathology or injury) during stair ascent and descent. Informed consent was obtained for each participant in this study. The group characteristics are as follows: mean age = 44.9 years ($\pm$ 9.48), mean height = 1.7 m ($\pm$ 0.09), and mean weight = 76 kg ($\pm$ 18.02).

Three-dimensional motion capture was performed using an 8 Qualisys ProReflex MCU 120 Hz digital camera system capturing at 60 Hz (Qualisys, Sweden). A custom staircase [**21**] was used, interfaced with a Bertec force plate (Bertec Corporation) capturing at 1080 Hz. The position of the force plate was defined relative to a global coordinate system (GCS) using the position data of markers attached to the corners of a panel positioned on top of the force plate. This enabled the centre of pressure coordinates of the ground reaction force to be expressed relative to the GCS.

Two marker sets were used simultaneously to allow the comparison of two methods of data collection and two methods of computing tibiofemoral kinematics.

Method 1 uses the approach of Holt *et al.* [**22**]. Figure 1 shows plate-mounted markers (a non-slip backing reduces slippage of the marker clusters relative to the skin) attached laterally to the thigh and shank. An anatomical calibration was performed with the subject in quiet standing. An aluminium pointer containing four retro-reflective markers was used to identify three bony landmarks per segment during 1 second recordings. These were the medial and lateral epicondylar gaps, medial malleolus, and the upper border of the greater trochanter. A 1 second static measurement with the subject in quiet standing was recorded prior to dynamic trials.

Method 2 uses commercial software, Visual3D (C-motion, Inc.). The landmarks from Method 1 were identified using passive markers attached to the skin. Additional markers were positioned in a modified Helen Hayes configuration (Fig. 2). A quiet standing

Fig. 1 The identification of a bony landmark (medial malleolus) using the pointer. Other bony landmarks identified during the calibration include the medial and lateral epicondylar gaps and the upper border of the greater trochanter

measurement was recorded with the subject's feet a shoulder width apart, for 1 s.

The stairs were constructed as shown in Fig. 3, with steps 1 and 2 individually in contact with a force plate. The subjects performed stair ascent and descent without the use of a handrail. Three trials of the following stair gait cycles (SGCs) of ascent and descent were recorded for each subject:

SGC1 (ascent) – right foot strike on step 1 through to right foot strike on step 3
SGC2 (ascent) – right foot strike on step 2 to right foot strike on 4
SGC3 (descent) – right foot off step 3 to right foot off step 1
SGC4 (descent) – right foot off step 4 to right foot off step 2

2.2 Data analysis

2.2.1 Knee kinematics

Method 1. Joint axes and rotations were defined according to the joint coordinate system (JCS) [**9**], following the recommendation by the International

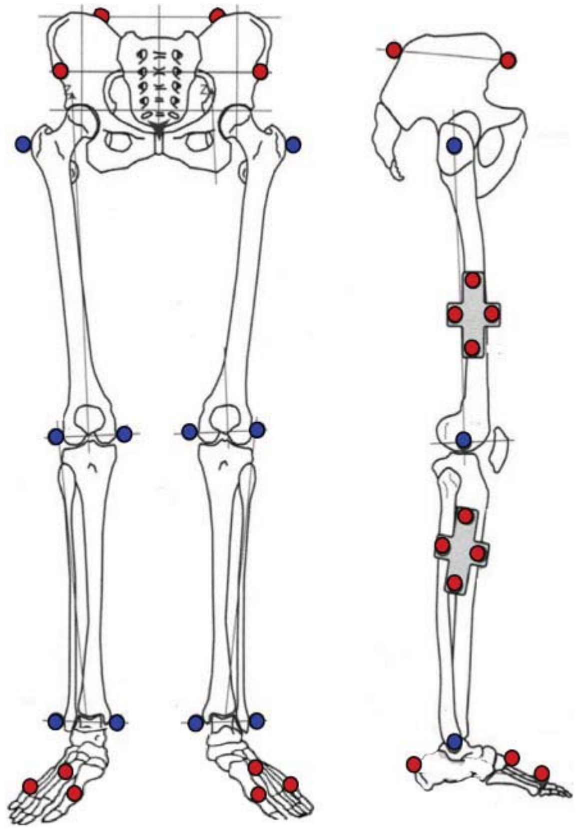

Fig. 2 Marker set showing the modified Helen Hayes configuration as in reference [**23**]

Society of Biomechanics (ISB) for standardization [**24**]. This was achieved using the method of Holt *et al.* [**22**] and in-house software (Matlab, Version 7.1, The Mathworks, Inc). This involved defining orthogonal axes in the femur and tibia using the pointer coordinates from the anatomical calibrations and the vector method. The origins of the axes were positioned midway between the femoral condyles. The static measurement was used to determine the relationship between the anatomical and technical axes in the marker clusters. Assuming rigid-body analysis, the position and pose of the segments were tracked using the rigid clusters of markers. The X axis was defined as the femoral flexion–extension axis, the Z axis was defined as the tibial internal–external rotation axis, and the axis orthogonal to the previous two at any instant in time was defined as the floating abduction–adduction axis.

Method 2. A lower limb biomechanical model was created for each subject from the static measurements using Visual3D (C-motion, Inc.). These were subsequently used for kinematic and kinetic analysis. The pose of each rigidly defined segment was determined by at least three non-collinear points using the vector method. The shank was defined

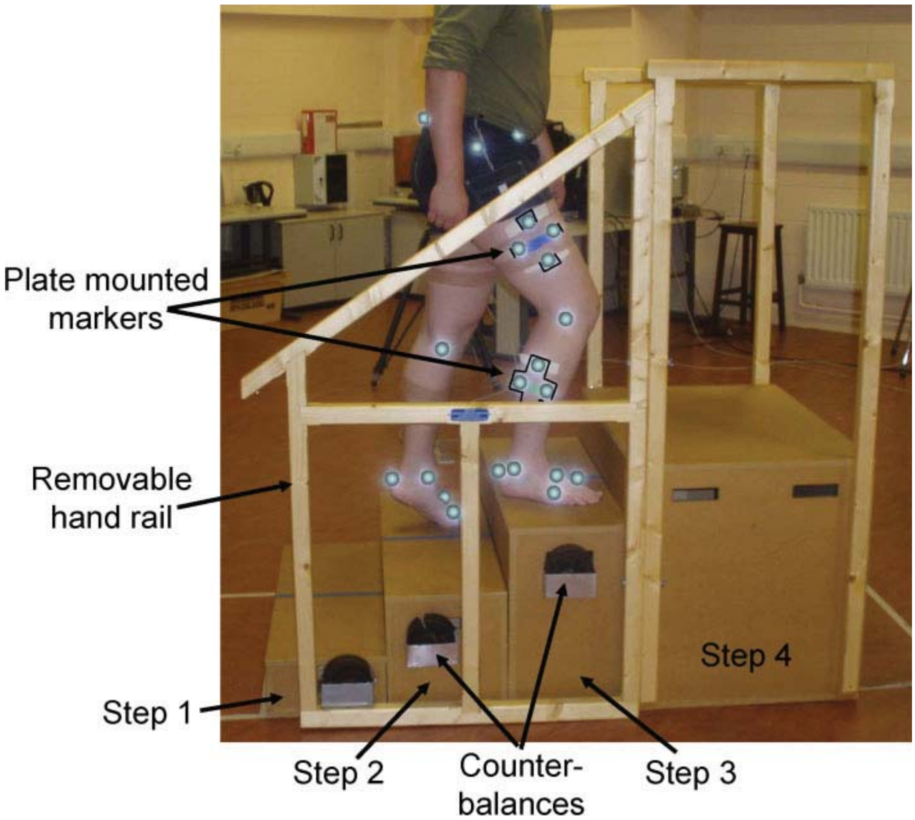

Fig. 3 Subject ascending from step 2 to step 3 of the staircase. Steps 1 and 2 can be positioned to interface with a force plate

using the position of the epicondyles and malleoli, and the thigh was defined using hip joint centre regression [**25**] and the epicondyles. The femoral axis system differed from that defined in Method 1 since the plane of the femur corresponds to the hip joint centre rather than the upper trochanter. Joint rotations were described by a Cardan–Euler sequence, where Z is the positive upwards vertical axis and Y is positive acting anteriorly. This is equivalent to the JCS [**9**].

For both methods, all rotation angles were defined by the orientation of the distal with respect to the proximal segment. An average of three trials for stair ascent and descent was computed for each subject. An unpaired independent t-test (SPSS 12.0.2) was performed to compare the kinematic measures from Methods 1 and 2.

2.2.2 Knee kinetics

Moments were described for the right leg relative to the laboratory GCS. The axes of the GCS are aligned such that the contributions of the moment acting about the x axis corresponds to the sagittal plane moment, about the y axis corresponds to the frontal plane moment, and about the z axis corresponds to the transverse plane moment. They are expressed as the contribution of the forces to rotate the shank about the knee joint centre, or 'external moments', and normalized to body mass. The outputs from three moment calculations are compared for each SGC where moments are computed from the stance phase of the SGC.

Moment calculation 1 (MC1). This method has been used in Andriacchi *et al.*'s study of stair climbing [**18**]. The moment of force is computed as the vector cross-product of a radius vector (position vector of the knee joint centre relative to the centre of pressure, or COP) and GRF vector using Matlab (Version 7.1, The Mathworks, Inc). The knee joint centre is computed from the position data collected using Method 1. Inertial effects were ignored as they are assumed to be small in low-velocity activities [**26**].

Moment calculation 2 (MC2). An inverse dynamics approach (IDA) was used to compute knee joint moments. The effect of the foot was ignored as no pointer position data were recorded for the foot segment. The mass, centre of mass (COM) position, and radius of gyration of the shank were determined using Zatsiorsky and Seluyanov [**27**]. The moment of inertia for the shank was determined from reference [17]. Segment accelerations were calculated using the kinematic data from Method 1.

Moment calculation 3 (MC3). Visual3D (C-motion, Inc.) was used to compute moments using a full IDA. This defines internal joint moments as the net internal moments generated by muscles crossing a joint. These were negated, converting them to external joint moments and normalized to body mass.

An average of the kinetic waveforms for three gait cycles was computed for each subject. Discrete parameters were extracted from the joint moment profiles for statistical analysis. One-way repeated measures of ANOVA were used to determine whether significant differences in these dependent variables occurred between the different computational approaches. For significant F ratios, a *post hoc* pairwise multiple comparisons Tukey test (SPSS 12.0.2) was performed.

3 RESULTS

3.1 Kinematics

The range of motion (ROM) of the kinematic waveforms and peak flexion angle computed from Method 1 and Method 2 are displayed in Table 2. Significant results were determined between the two computational methods for frontal ROM for SGC2 of ascent and both SGC3 and SGC4 for descent. Examples of the joint kinematic waveforms and the discrete peak values used for comparison are given in Table 2 and displayed in Fig. 4.

3.2 Kinetics

Discrete values from the moment profiles are displayed in Table 3. Significant differences in the joint moment profiles were found when using MC3 as compared to MC1 and MC2 for each SGC of stair ascent and descent. Forces were measured from step 1 for SGC1 and SGC3 and step 2 for SGC2 and SGC4. Examples of the joint moment profiles and the discrete peak values used for comparison are given in Table 3 and displayed in Fig. 5.

4 DISCUSSION

The kinematic and joint moment profiles are consistent with previous studies [**1**, **12**, **13**, **15**, **16**, **18**]. The kinematic waveforms followed the same patterns for Methods 1 and 2. In a comparison of discrete variables from the waveforms, a significantly larger frontal ROM was calculated using Method 1 for stair ascent for SGC2 and for both SGCs of stair descent. A larger frontal ROM was also noted for SGC1, but, due to a large variability in the data, this result was not significant. The adduction–abduction axis used to compute rotations in the frontal plane is determined as a cross-product of the two vectors defined by the anatomical landmarks. A difference in the position of these landmarks affects the anatomical coordinate system in the femur, which may have an effect on small rotations in the frontal plane. The thigh segment is defined differently for Methods 1 and 2. Method 1 uses the greater trochanter for the proximal landmark, whereas Method 2 uses the hip joint centre, producing different alignments of the

Table 2 Kinematic measures used to compare Method 1 and Method 2

	Variables (deg)	Method 1 ($n = 10$)	Method 2 ($n = 10$)
Ascent step 1 to step 3 (SGC1)	Sagittal ROM	76.82 ± 3.19	77.75 ± 4.30
	Peak flexion angle	85.47 ± 5.89	87.67 ± 5.06
	Frontal ROM	16.41 ± 8.18	10.89 ± 2.92
	Transverse ROM	14.70 ± 4.05	12.73 ± 3.51
Ascent step 2 to step 4 (SGC2)	Sagittal ROM	80.27 ± 6.33	80.79 ± 7.97
	Peak flexion angle	87.73 ± 7.20	89.73 ± 6.59
	Frontal ROM	17.50 ± 8.07	*11.25 ± 2.82
	Transverse ROM	15.78 ± 3.62	13.68 ± 4.21
Descent step 3 to step 1 (SGC3)	Sagittal ROM	81.16 ± 6.63	80.26 ± 5.61
	Peak flexion angle	88.58 ± 8.31	88.09 ± 6.56
	Frontal ROM	19.45 ± 7.84	*8.87 ± 1.96
	Transverse ROM	15.02 ± 5.93	11.97 ± 3.72
Descent step 4 to step 2 (SGC4)	Sagittal ROM	81.40 ± 6.86	80.11 ± 6.52
	Peak flexion angle	89.49 ± 9.13	89.14 ± 7.87
	Frontal ROM	19.57 ± 7.20	*9.89 ± 1.20
	Transverse ROM	15.65 ± 5.37	13.94 ± 6.70

Mean ± standard deviation; *indicates a statistical significance between the data collection methods ($p < 0.05$).

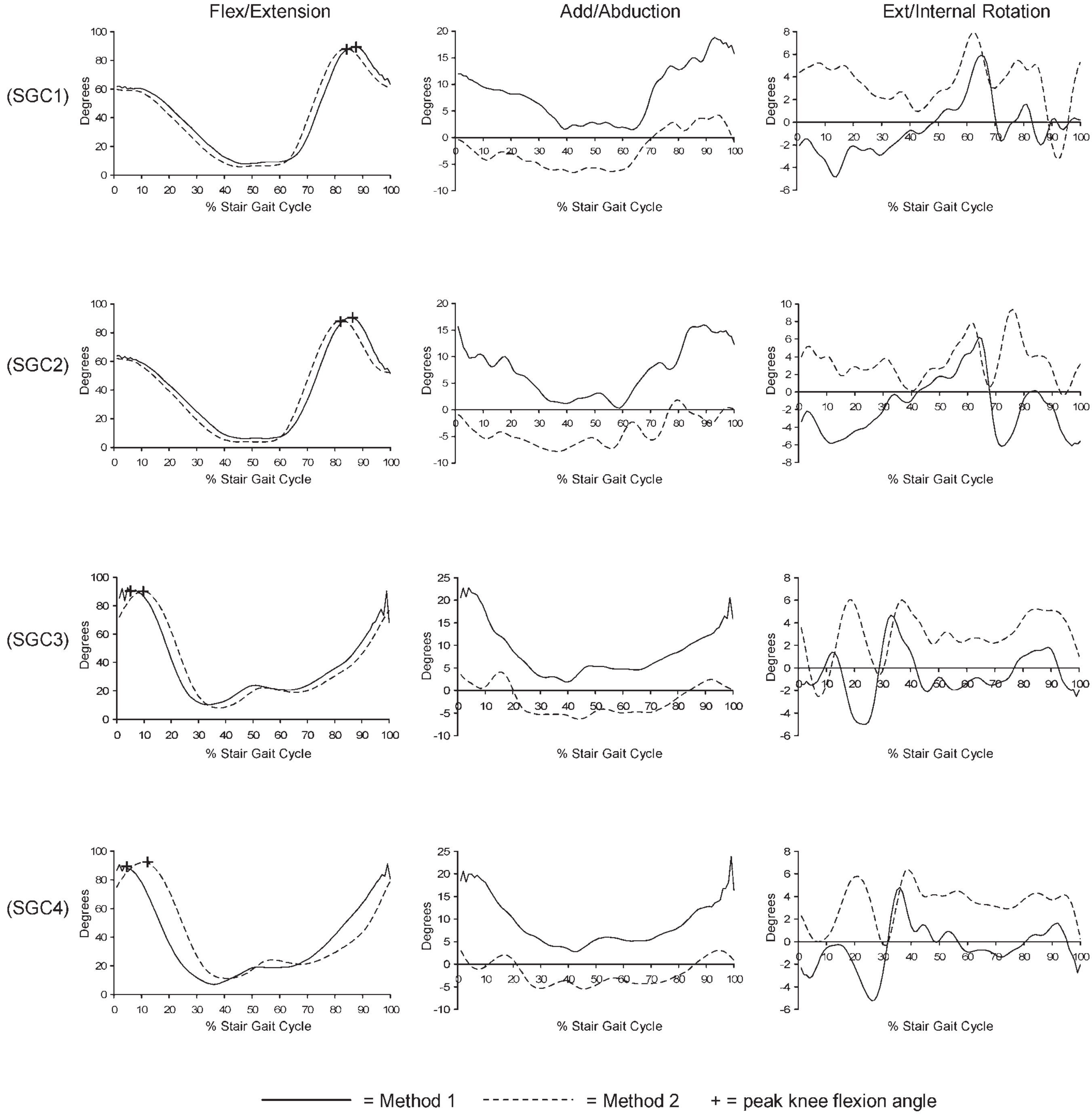

Fig. 4 Examples of knee kinematic waveforms for: (SGC1) stair ascent from step 1 to step 3; (SGC2) stair ascent from step 2 to step 4; (SGC3) stair descent from step 3 to step 1; (SGC4) stair descent from step 4 to step 2. Each waveform represents a mean of three trials from a single subject. The solid line represents the kinematics from Method 1; the dashed line represents the kinematics from Method 2; + indicates the peak flexion angle

thigh segment axis. The variation in frontal ROM was greater from Method 1 compared with Method 2. This may be due to Visual3D (C-motion, Inc.) using an optimal method for tracking segments.

Three methods for computing knee joint moments were explored. MC1 is a robust method of computing net moments about a joint without requiring knowledge of individual moments acting about the joint. Including inertial effects of the shank, as in calculation MC2, did not have any significant effects. MC2 could have been adapted to the method of reference [**12**] where the moment of inertia is

Table 3 Kinetic measures used to compare MC1, MC2, and MC3

	Variables (N m/kg)	MC1 ($n = 10$)	MC2 ($n = 10$)	MC3 ($n = 10$)	p value
Ascent step 1 to step 3 (SGC1)	Peak flexion moment	1.19 ± 0.24	1.17 ± 0.23	0.86 ± 0.18	0.003*
	Peak extension moment	0.42 ± 0.18	0.45 ± 0.19	0.46 ± 0.17	0.859
	Peak adduction moment	0.32 ± 0.10	0.30 ± 0.10	0.30 ± 0.08	0.845
	Peak external rotation moment	0.06 ± 0.02	0.06 ± 0.02	0.07 ± 0.02	0.304
	Peak internal rotation moment	0.02 ± 0.12	0.02 ± 0.01	0.05 ± 0.02	0.000*
Ascent step 2 to step 4 (SGC2)	Peak flexion moment	1.21 ± 0.26	1.20 ± 0.26	0.79 ± 0.21	0.001*
	Peak extension moment	0.47 ± 0.15	0.49 ± 0.16	0.44 ± 0.11	0.754
	Peak adduction moment	0.36 ± 0.10	0.34 ± 0.10	0.23 ± 0.09	0.007*
	Peak external rotation moment	0.05 ± 0.02	0.05 ± 0.02	0.06 ± 0.02	0.403
	Peak internal rotation moment	0.01 ± 0.01	0.02 ± 0.01	0.05 ± 0.02	0.000*
Descent step 3 to step 1 (SGC3)	Flexion moment peak 1	0.66 ± 0.33	0.67 ± 0.34	0.57 ± 0.29	0.754
	Flexion moment peak 2	0.83 ± 0.21	0.78 ± 0.20	0.97 ± 0.19	0.088
	Adduction moment peak 1	0.40 ± 0.15	0.38 ± 0.15	0.21 ± 0.18	0.026*
	Adduction moment peak 2	0.39 ± 0.18	0.37 ± 0.18	0.22 ± 0.17	0.060
	Peak external rotation moment	0.10 ± 0.27	0.10 ± 0.03	0.13 ± 0.04	0.123
	Peak internal rotation moment	0.02 ± 0.01	0.02 ± 0.01	0.02 ± 0.02	0.546
Descent step 4 to step 2 (SGC4)	Flexion moment peak 1	0.74 ± 0.36	0.73 ± 0.36	0.60 ± 0.34	0.633
	Flexion moment peak 2	0.86 ± 0.21	0.82 ± 0.21	1.17 ± 0.25	0.003*
	Adduction moment peak 1	0.57 ± 0.14	0.55 ± 0.14	0.28 ± 0.15	0.000*
	Adduction moment peak 2	0.48 ± 0.14	0.45 ± 0.14	0.21 ± 0.13	0.000*
	Peak external rotation moment	0.10 ± 0.01	0.10 ± 0.02	0.11 ± 0.02	0.178
	Peak internal rotation moment	0.02 ± 0.01	0.02 ± 0.01	0.02 ± 0.01	0.391

Mean ± standard deviation; *significant comparisons ($p < 0.05$) for MC3 versus MC1 and MC2.

modified to account for foot mass, but considering the similarity of the results from MC1 and MC2 it is unlikely to have a large effect. The only significant differences were found between the IDA approach MC3 compared with MC1 and MC2. The movement of the foot appears to have a greater effect on the resulting moments as compared to including the inertia of the shank alone. This could be attributable to the inertia of the foot generating greater moments about the knee because it is further away or to the different methods of data collection. Also, inertial effects modify the acceleration and deceleration of the whole body COM, producing different GRF curves influencing the full IDA method MC3.

The transverse moment is not widely reported in studies. As only two significant differences in transverse moment were found between the data collection methods in this study, it is questionable as to whether in future studies the transverse moment should be more readily used.

This study highlights the implications of comparing data from different analysis methods. Clearly describing data collection and analysis methods will enable educated judgements to be made when interpreting and comparing results from different studies. A wide range of limb configurations are mechanically feasible during stair ascent/descent [**28**]. Moment profiles from previous studies display different patterns due to methodology, even though the magnitudes are comparable [**13**]. This can be seen for the adduction–abduction and external moment profiles in Fig. 5. This study has shown that differences in the kinematic and kinetic outcomes can occur due to the assessment and analysis methods used. These differences would be critical if methodologies were to change over the course of a long-term study.

This work recognizes the benefits of developing standards for the assessment of activities where methodology has a significant effect on biomechanical outcomes. It is important when comparing outcomes from a range of studies to identify the differences that exist solely due to the varying strategies adopted for stair gait for healthy, pathology-related, or rehabilitation regimes. This would remove the need to discern differences that are clouded by the disparity that arises when employing varying measurement and computational methods. In order to develop a standard to allow direct comparison of cross-laboratory data, a larger population of subjects must be recruited, which raises the idea of a larger cross-centre study.

In future work, beyond the scope of this paper, consideration should also be given to the reference frames for the expression of moments. For this study the orthogonal laboratory GCS was used. Alternative orthogonal frames are the proximal segment coordinate system and distal segment coordinate system. Another possibility is the non-orthogonal JCS. There is no consensus regarding an accepted standard and this contributes to the difficulties in comparing joint moment data across studies.

It is an accepted standard to define lower limb rotations according to the non-orthogonal JCS for the

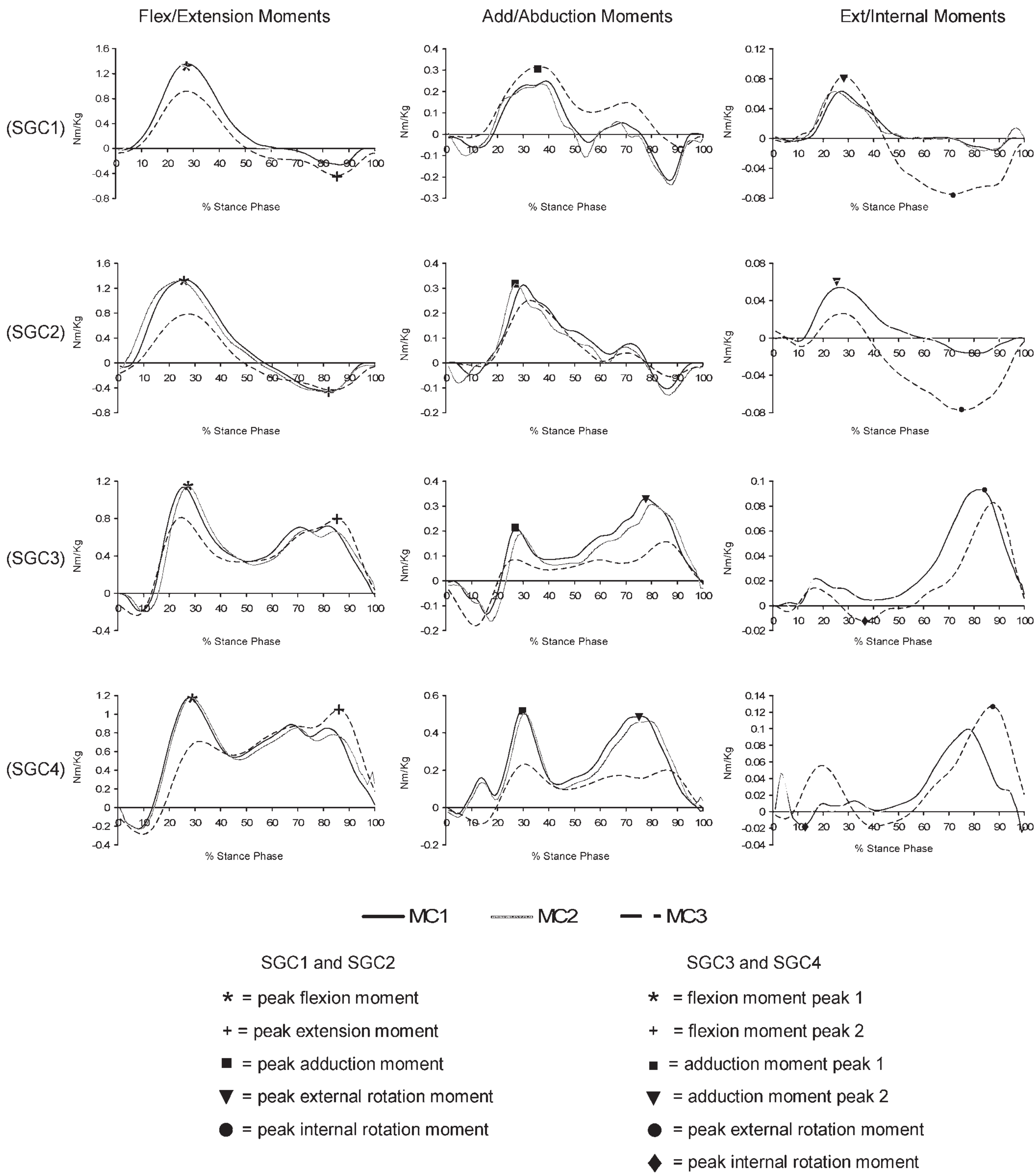

Fig. 5 Examples of knee moment profiles for a single trial of: (SGC1) stair ascent from step 1 to step 3; (SGC2) stair ascent from step 2 to step 4; (SGC3) stair descent from step 3 to step 1; (SGC4) stair descent from step 4 to step 2. For clarity, each discrete peak value is indicated for an individual waveform as an example. The solid line represents the knee moment computed using MC1; the dot–dashed line represents the knee moment computed using MC2; and the dashed line represents the knee moment computed using MC3

clinical interpretation of joint moments. It has been suggested that all calculations should be conducted in an orthogonal reference frame and then converted to the non-orthogonal frame for interpretation [**29**]. In a future study it would be beneficial to identify the influence of the reference frame used for the computation of knee joint moments during stair gait. In the study by Schache and Baker [**29**], significant differences in the joint moment profiles during level gait were found with alternative reference frames and it is hypothesized that these differences would be amplified when considering stair gait.

From this investigation, although the full IDA approach MC3 utilizes information on lower limb segment properties, which leads to a more informative solution for joint moment calculations, this requires information in the foot segment properties that is not available using the current Cardiff protocol employing the pointer method of anatomical calibration. Moment calculation MC1 has been used successfully in previous studies and although the computation is basic compared with the IDA approach, it can be utilized with the pointer method of computing kinematics. It is a method where the directions of individual forces do not need to be known as with the IDA method.

Regardless of the computation method, moment data are interpreted using a form of pattern recognition based on deviations of signals from a normative equivalent. For this reason, if the marker set does not allow the computation of joint moment using IDA, as long as the limitations of this approach to computing moments are recognized and results are interpreted accordingly, meaningful data can be obtained.

This study highlights the differences in kinematic and kinetic data that can result from the use of different data collection and analysis methods. It also raises the idea of standardization to allow direct comparison of cross-laboratory data. This would require a larger cross-centre study and larger cohorts to investigate a broad range of methodologies.

ACKNOWLEDGEMENTS

Funding from DePuy International Ltd supported this study. Thanks are given to Dr Lianne Jones for assistance with data collection. Dr Jones was a postdoctoral RA at the Cardiff School of Engineering, Cardiff University, and is now teaching Physics in Llantrisant, South Wales.

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