University of Ottawa

Assessment of Functional Knee Bracing: An *In-vivo* Three-dimensional Kinematic Analysis of the Anterior Cruciate Deficient Knee

by

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Thesis

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Abbreviations & Definitions

ACL
anterior cruciate ligament

ACLD
anterior cruciate ligament deficient

Anterior drawer
Hip flexed 45° with knee at 90°. Firm pressure is applied to the posterior tibia in an effort to translate it anteriorly.

ATT
anterior tibial translation

Cross cut
Defined as cut with which the tested leg is planted and the other leg crossed over to make the cut toward the side of the tested leg. (simulates pivot shift)

DLT
direct linear transform

EMG
electromyography or neuromuscular activity

FI
Functional instability. Subjective symptoms presented by the individual such as feeling of instability and or recurrent symptomatic subluxations

GRF
ground reaction forces

ILED
inter LED distance

IT band
Iliotibial band

Joint co-ordinate system
A co-ordinate system in which there are two body-fixed axes with the third axis being perpendicular to the other two. In the knee, the femoral fixed axis is the mediolateral axis which passes through the origin of the femoral anatomical system at the floor of the inter-trochanteric groove. Positive is in the lateral direction. The tibial fixed axis is the longitudinal axis which passes through the origin of the tibial anatomical system which courses between the medial and lateral tibial eminence and is at the level of the highest one. The third axis, the antero-posterior or floating axis is formed by the common perpendicular to the fixed femoral and tibial axes.

Lachman's test

Clinical measurement of anterior sagittal movement of the tibia in relation to the femur in 15°-20° of knee flexion. The femur is stabilized and firm pressure is applied to the posterior tibia in an effort to translate it anteriorly.

Losee test

Test anterior rotary subluxation of the lateral tibial plateau. The foot and ankle are externally rotated with the knee in 30% of flexion. A valgus stress is applied and the head of the fibula is pushed anteriorly while allowing the knee to sink into extension. If the lateral tibial plateau subluxes anteriorly and the patient recognizes the movement as the cause of the disability, the test is positive.

OLH

one legged hop

PFP

patellofemoral pain

Pivot shift

Clinical measurement of anterior subluxation of the tibia in relation to the femur. In the supine position with the knee extended and the foot internally rotated, a valgus stress is applied to the knee. As the knee is flexed, the tibial plateau will reduce with a shift at 20°-40° of knee flexion if anterior subluxation is present.

RSA

roentgen-stereo-photogrammetry (rorentgen stereo analysis)

SL

static laxity. An increased SL is defined as mobility beyond physiological limits due to a capsulo-ligamentous injury

SMAC

simultaneous multiframe analytic calibration

Torque

The turning effect produced by a force. Calculated as the product of the force and the perpendicular distance between the point of application of the force and the axis of rotation

3D

three-dimensional

Valgus

The condition of outward deviation in alignment from the proximal to the distal end of a body segment.

Varus

The condition of inward deviation in alignment from the proximal to the distal end of a body segment.

Acknowledgements

Success is not a measure of achievement, but is a measure of what we overcome.

I would like to extend my appreciation to a number of people. First, I would like to express my sincere gratitude to Dr. Mario Lamontagne (University of Ottawa, Canada) and Dr. Gunnar Németh (Karolinska Hospital, Sweden) for providing me the opportunity to conduct my research overseas. Many graduate students only dream of being able to conduct their research abroad while experiencing a new culture and the camaraderie of fellow foreign students. For my experience, I am truly grateful and hopeful that my research will foster continued collaboration.

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Assessment of Functional Knee Bracing: An In-vivo Three-dimensional Kinematic Analysis of the Anterior Cruciate Deficient Knee

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Abstract

The aim of this investigation was to determine whether application of a functional brace reduced rotational and linear tibial displacements during the performance of a *One Legged Jump* (OLJ). Steinmann traction pins were surgically implanted into the femur and tibia of six young normal healthy subjects having either a partial or complete anterior cruciate ligament (ACL) rupture. Stereophotogrammetric radiographs (RSA) were taken once target markers were affixed to the pins. Angular and translation measurements were recorded using the MacReflex motion analysis system sampling at 120 Hz. A Kistler force plate was synchronised to collect ground reaction forces simultaneously at 960 Hz. Patients were required to jump for distance to sufficiently stress the ACL. Subjects were randomly assigned to start with either the braced or unbraced condition. Analysis focused on differences in magnitudes and changes in the shape of the curves between bracing conditions. Intra-subject peak vertical force and anterior shear force was generally consistent between unsupported and braced conditions; indicating jumps onto the force platform were similar. The small intra-subject angular and translational differences cannot be attributed to variations in jumping styles, but rather to the brace itself. Magnitude's varied across subjects since subjects jumped within their own comfort limits. Tibiofemoral rotations and translations show a general trend across subjects, i.e., the shape and amplitudes of the skeletal marker based curves were similar. The major difference is a shift between the unbraced and supported conditions. The offset between conditions can be attributed to the brace or the different standing reference trials used during the unbraced and braced trials. This created small differences in alignment of the tibial and femoral anatomical coordinate systems rather than to application of the brace itself. Generally, intra-subject knee kinematics were very repeatable but differences between unbraced and braced patterns were small. This may be due to the

invasiveness of this protocol, that landings are performed onto a deficient limb, and that subjects jumped within their own comfort limits which did not maximally stress the ACL. As expected, inter-subject differences were typically much larger than intra-subject variability. Differences mainly consisted in amplitudes and position at touchdown.

Introduction

The criterion for determining whether ACL reconstructive surgery is required is based on the patients' functional instability that is determined from physical and instrumented tests (e.g. KT1000). Alternatively, functional knee braces have been designed to stabilise deficient knees by reducing pathological translations and rotations. Yet little research has examined the effects of knee braces on three-dimensional osteokinematics and arthrokinetics during high physiologic conditions. Braces are effective in reducing anterior translations when subjected to static or low anterior shear forces but fail in situations where high loads are encountered or when the load is applied in an unpredictable manner [1,2,3,4,5,6].

Knowledge about skeletal tibiofemoral joint motion is limited, in particular the secondary rotations and linear translations. Although recent investigations have used invasive markers to directly measure tibiofemoral joint motion, these studies have been restricted to semi-static activities, or walking and light running [7,8,9,10,11]. Since braces are designed for athletic activity, they should be evaluated under such conditions. Therefore there is a need to quantify true anatomical tibiofemoral motion during strenuous activity.

This study involved new techniques including intracortical pin implantation and 3D-motion analysis to assess tibiofemoral joint kinematics for anterior cruciate ligament deficient (ACLD) knees during a functional task. The aim of this investigation was to determine whether application of a functional brace reduced rotational and linear tibial displacements during the performance of a *One Legged Jump* (OLJ).

Material and Methods

Subjects

Six male subjects with ACL deficient knees and having no prior surgical treatment (age 21.8 ± 4.17 years, mass 80.83 ± 8.23 kg, height 181.50 ± 6.92 cm) were selected by an orthopaedic surgeon to participate in the study. Each had a history of significant instability that caused them to modify their activity. Patient's knees exhibited at least a +3 laxity score compared to their contralateral leg when evaluated with the KT 1000 arthrometer. Participants signed an informed consent form to participate in the study. The Ethics Committee of the Karolinska Hospital approved the experimental procedure and the surgery was performed at the Department of Orthopaedics, Karolinska Institute. Data collection was performed at the Motor Control Laboratories located at S:T Göran's Hospital and Astrid Lindgren Childrens Hospital, Sweden.

Surgical procedure

Intracortical Steinmann bone pins (2.5 mm diameter) were inserted with a manual orthopaedic drill into each of the subject's deficient leg. Prior to insertion, the skin, subcutaneous tissue and periosteum were anaesthetised with standard anaesthetic. To minimise impingement problems with the iliotibial band, the knee was flexed 45° prior to pin implantation [9]. The pins were inserted anterolaterally and superior to the femoral condyle and antrolaterally into the proximal portion of the tibia. These insertion sites ensured no impingement occurred between the brace and Steinmann pins during the dynamic functional task. Target clusters were then affixed to the pins (Figure 1). Each target marker was comprised of four non-collinear 7 mm reflective markers, one in the centre and three attached to orthogonal projecting rods [8,9,10,11]. Since the anaesthetic was generally active for 2 hours, this left ample time for the motion recordings. The pins remained inserted for the duration of the test.

Roentgen-stereophotogrammetric x-rays were taken with the implanted pins to record the position of the markers and to define the tibial and femoral anatomical reference points. All radiographs were taken with the subject supine on the x-ray table with the leg extended and flexed approximately 10° [9]. The anatomical axes were defined as follows (Figure 2): The Z axis was oriented in the vertical direction and ran parallel to the nominal longitudinal axes of the femur and tibia; the X axis in the anteroposterior direction and perpendicular to the Z axis; and the Y axis mediolateral and mutually orthogonal to both "Z" and "X" axes [8]. For the femoral coordinate system, the deepest point of the intercondylar groove was chosen as the origin. The Z_f axis passed through the origin and was directed superiorly and parallel to the femoral longitudinal axes. In the frontal plane, the Z_f axis was positioned medially from the femoral bisector while in the sagittal plane it was posterior. The Y_f axis was perpendicular to the Z_f axis and directed from the lateral condyle to the medial condyle while passing through the origin. The

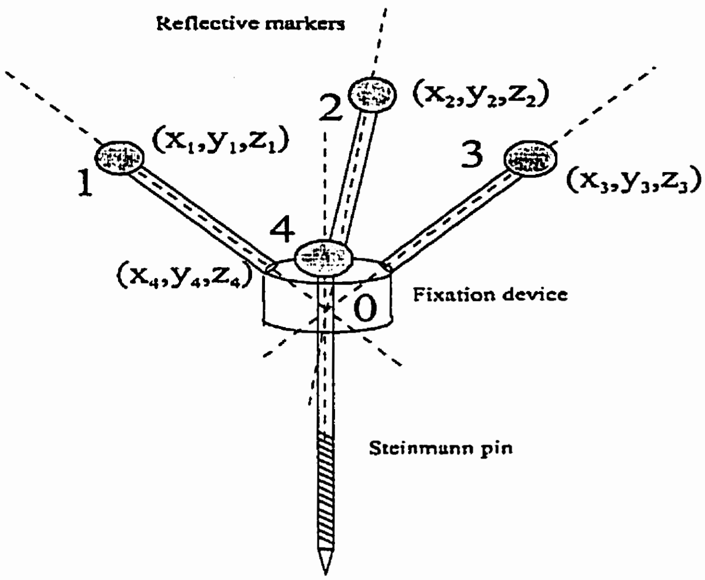

Figure 1: Schematic of target cluster (Adapted from Lafortune, 1984, pp. 67)

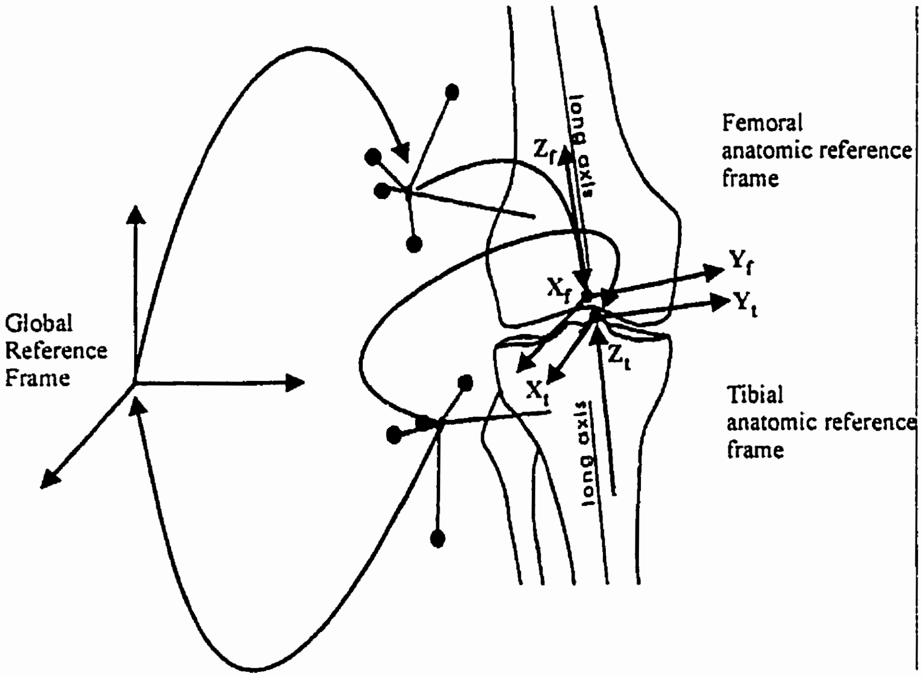

Figure 2: **Anatomical reference frame for the femur and tibia**

mutually orthogonal X_f axis progressed from the posterior to the anterior femur passing through the origin. The origin for the tibial coordinate system was located on the most proximal point of the medial intercondylar eminence. The tibial Z_t axis ran parallel to the longitudinal axis of the bone, was directed superiorly and passed through the origin. Additionally, the Z_t axis was medial and posterior to the tibial bisectors in the frontal and sagittal plane respectively. The Y_t axis was orthogonal to the Z_t axis and progressed from the lateral to the medial tibial articular surfaces while passing through the origin. The X_t axis was mutually orthogonal to both Z_t and Y_t axes and directed from posterior to anterior passing through the origin, which in the sagittal plane was just anterior to the tibial eminence [8,9,10,11].

Motion recordings

Six MacReflex infrared 60 Hz cameras were paired and affixed to specially designed tripods to record the motion. The MacReflex motion analysis system was synchronised so that the two 60 Hz cameras in each pair recorded in alternate frame sequences, equivalent to three twin cameras sampling at 120 Hz. Each camera was equipped with lenses to give a horizontal field-of-view of 28°. Prior to recording, a calibration frame with nine control points (volume 25 $\times$ 49 $\times$ 15 cm^3) was used to calibrate the measurement area approximately 45 cm off the floor (representative of knee height). Camera pairs were orientated to obtain a field of view covering the entire dimension of the calibration grid. Target markers were visible in all cameras during stance. Ground reaction forces were simultaneously collected with a Kistler force plate sampled at 960 Hz. Motion recordings and the force platform were synchronised with an external trigger to collect simultaneously upon commencement of the jumping manoeuvre.

Following calibrations, the subject was aligned so that the sagittal plane was oriented with the X-Z plane of the MacReflex-calibrated system (with the Z axis directed vertically). A standing reference trial was recorded with the subject in this controlled posture. (Figure 3). Two additional 7 mm markers were placed on the corners of the force platform to set the correct aperture for the MacReflex cameras for each motion recording.

Experimental protocol and set-up

Prior to surgery, patients completed the Lysholm Knee Scoring Scale [12] to assess their loss of knee function and the Tegner Activity Grading Scale [13] that ranks activities according to how troublesome they are to perform. Their activity levels were later analysed in relation to the Lysholm Knee Scoring Scale. After pin implantation, each subject was given several trials to perform the *One Legged Jump* (OLJ) to familiarise themselves with the pins and testing protocol. To sufficiently stress the ACL, each subject jumped for maximal horizontal distance. From an initial standing position with the deficient limb set back, the subject pushed off from their sound limb and landed with their deficient limb. Their longest measurement was recorded and marked on the floor to determine the proper takeoff distance to the force platform. After familiarisation with the procedure, two standing reference trials and five measurement trials were recorded. Standing reference trials were recorded prior to and following the functional tasks. For the standing trial, subjects stood in a neutral position and were instructed to align their feet parallel to the force platform to define the tibial and femoral anatomical coordinate system. It was arbitrarily defined that the segmental coordinate systems were aligned with the global coordinate system during standing.

Each subject was tested during a single experiment session, wearing their own running shoes and dark lightweight clothing for ease in identifying markers. Subjects were randomly assigned to start with either the braced or unbraced condition. Placement of the brace (DonJoy Legend) was applied by the researcher according to the specifications prescribed by the manufacturer. After the standing trials and five measurement trials were completed for the first test condition, two additional standing trials and five measurement trials were collected for the subsequent test condition. Synchronisation between the jump and data collection was initiated with a verbal que. Having given the command to start, data collection and the performance of the jump commenced.

Three-dimensional reconstruction

For each camera pair and subject, both the standing and measurement trials were manually sorted and autotracked using MacReflex 3.2 PPC data acquisition software. Autotracking transformed the 2-D image coordinates onto a 3D coordinate system employing MacReflex's Direct Linear Transform (DLT) [14] algorithms. All cameras were used for the three-dimensional reconstruction. Incorrect markers were invalidated and any missing or hidden markers were filled using linear interpolation. After autotracking, the data including the frame numbers were exported so the output files from the *Segmental Analysis* calculations (The

Lundberg Laboratory for Motion Analysis, Göteborg, Sweden) corresponded with the original MacReflex recording.

Reference frames and relative orientation

The *Segmental Analysis* software calculated relative 3-D motions between the anatomical tibial reference frame relative to the femoral anatomical reference frame. Results were expressed as either relative 3-D angular orientation or relative 3-D displacements between the two fictive points. Angular descriptions (between the distal and proximal segment) were described using the conventions of Kadaba [15] and Davis [16] and computed using the rotation sequence about –y, x, z axes [17]. The *alpha* angle is rotation of the measurement segment (tibia) in the X-Z plane of the reference segment (femur). The alpha rotation is positive about an axis parallel to the negative Y axis of the reference system (Figure 4). A positive value indicates hyperextension of the knee. The *beta* angle is rotation of the measurement segment in the Y-Z plane of the reference segment. The beta-rotation is positive about an axis parallel to the positive X-axis of the reference system. A positive beta angle about the floating axis indicates *adduction* of the lower leg in relation to the thigh (foot brought in towards the midline of the body). Conversely, a negative value indicates *abduction*. The *gamma* angle is rotation of the measurement segment in the X-Y plane of the reference segment. The gamma-rotation is positive about an axis parallel to the positive Z-axis of the reference system. A positive *gamma* angle represents internal rotation of the lower leg (tibial tubercle towards the midline of the body) whereas a negative value indicates external rotation. Based on a right-handed coordinate system, for the right leg the Y-axis was directed medially whereas for left leg it pointed laterally. The difference in orientation between limbs was accounted for by manually negating the Y coordinates in the left leg and utilising the left handed coordinate system to describe rotations.

The anatomical coordinate system was defined using each patients neutral standing trial. During standing, subjects aligned their segments with the force plate and X-Z plane of the MacReflex system which represented the global coordinate system (Figure 3). It was arbitrarily defined that the anatomical coordinate systems were aligned with the global coordinate system. A coordinate transformation matrix (derived from three rotational and three translational degrees of freedom) resolved the tibial anatomical reference frame into the femoral anatomical reference frame. A set of three independent angles and translations were extracted by decomposition and

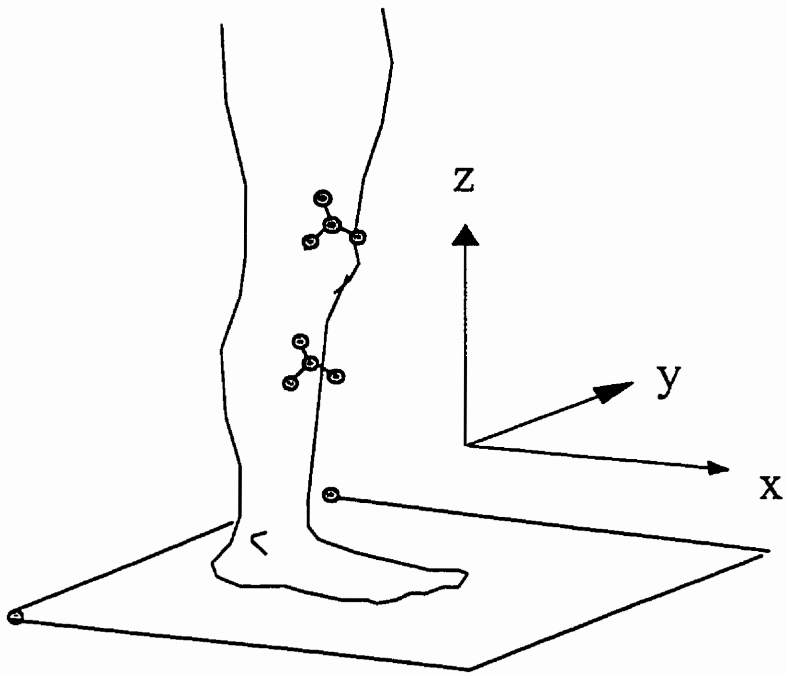

Figure 3: Orientation of foot on the force platform with target clusters attached to the thigh and lower leg. (adapted from *Segment Analysis* manual © Karlsson, 1997 pp 9)

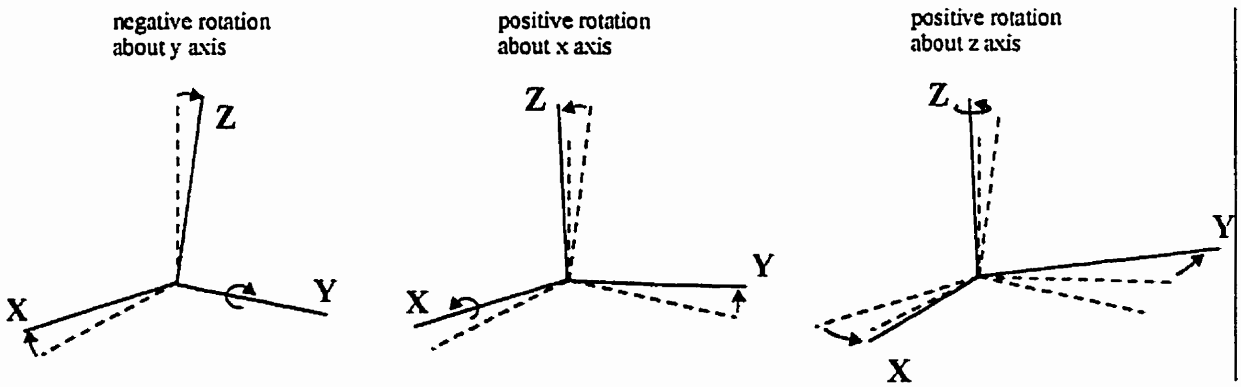

Figure 4: Angular descriptions employed by *Segment Analysis* (The Lundberg Laboratory for Motion Analysis, Göteborg, Sweden Karlsson et al. 1994)

normalised to the standing reference trial to in order to describe movement of the segmental (anatomical) coordinate systems [18]. The methods used to calculate the transformation matrices are reported in greater detail elsewhere [8,9,18,19,20,21,22]. Tibiofemoral joint motion was described using Grood and Suntay's joint coordinate" system [23]. General joint motion was partitioned into six familiar anatomic motions employing Cardan angles. According to the conventions described by Grood and Suntay [23], flexion/extension and medial-lateral shift occurred around the fixed medio/lateral femoral axis, ab/adduction and anterior-posterior drawer around the floating axis and internal/external knee rotation and distraction-compression around fixed tibia proximal/distal axis.

Kinematic data derived from the Segment Analysis software were filtered with a Butterworth 4th order, low-pass, critically damped, zero-lag filter with a cutoff frequency of 6 Hz. The cutoff frequency was determined by running a Fourrier analysis of the angular and translational data respectively and by visual inspection. Additionally, jumps for each subject and condition were time normalised to 100% and an average derived. Because of the variability of the jumps across subjects, both kinematic and kinetic data was normalised to specified time intervals. Briefly, the point when HS occurred was obtained from the force platform data and the corresponding frame number identified in the kinematic data. The landing-stance cycle was calculated 50 ms prior to HS to a point when peak extension occurred and the associated posterior ground reaction shear force reached a plateau. All relevant force data were associated with the coincident kinematic frame number.

Peak vertical load and anterior posterior shear forces were scaled to body weight (including the brace when applicable) and interpolated so that each body position during the landing had a corresponding applied ground reaction force. Additionally, ground reaction forces were time normalised to 100% for each subject and condition using the same time criteria established for the MacReflex (kinematic) data. Initial contact with the force platform was noted to coordinate film and GRF data. If peak vertical forces were similar for both jumping conditions, the differences in translational data may be attributed to the brace rather than differences in jumping.

Assumptions and limitations

Motion recordings of skeletal movement employing intracortical pins is an invasive procedure. This may cause discomfort or the anaesthetics may alter the subject's perception.

Previous investigators have reported subjects did not experience significant discomfort, they moved their knees freely and walking and running styles remained unaffected [9,10,20,22]. However, due to the invasiveness of this investigation and since subject's jump onto their deficient limb, jumps were within the patient's comfort limits. This may not be adequate to maximally stress the ACL.

The MacReflex calibration frame required to calibrate the measurement area was limited due to the insufficient number of calibration points (nine). The accuracy of spatial reconstruction is reduced when a small number of calibration points are used [24]. However, during all motion recordings, the markers remained within the calibrated volume.

Results

Preceding target marker fixation and RSA x-rays, subjects performed moderate flexion/extension manoeuvres to assess possible impingement problems between the iliotibial band and femoral pin. Problems were encountered with subject 2 as the femoral pin bent during flexion. When the knee under went large flexion angles, the interaction of the soft tissue, musculature and iliotibial band generated enough force to bend the femoral pin approximately 10° (about the long axis). The pin was surgically removed and the subject was excluded from the study. For the remaining subjects, larger incisions were made about the femoral insertion site and flexion angles were restricted. No subjects experienced significant discomfort and they reported they could move their knees freely. Additionally, subject 5 was excluded due to significant marker dropout in the kinematic data rendering linear interpolation impossible.

During the experiment, none of the subjects reported their ability to jump was affected by the pins. It must be noted subjects performed the *One Legged Jump* (OLH) within their own comfort limits. The Lysholm Knee Score averaged across six subjects was 72.5 (2.6) ranging from 69 to 75. The mean Activity Score was 6.0 (1.9) and ranged from 4 to 9. All subjects had difficulty in sport and is reflected by the low Activity Score. None of the subjects had difficulty during daily activity as indicated by the moderate Lysholm Knee Score. Following peak flexion after the landing, subjects either began to extend, remain flexed or further flexed following a brief stabilisation period.

Due to the narrow field-of-view of the motion analysis system and the close proximity of the cameras about the knee, recording a full cycle from foot-strike to toe-off was not possible.

All the markers came into view at about foot-strike until the knee began to extend. Both angular and linear tibiofemoral joint patterns are illustrated in Figures 5 and 6. Standard deviations less than 0.6° for rotations and translations less than 0.4 mm have been reported when comparing RSA values and MacReflex data recorded in a volume of 0.25 m^3 [25].

Each subject served as their own control with analysis focusing on differences in magnitudes and changes in the shape of the curves between conditions and across subjects. Average curves were derived using trials collected for each of four subjects during unbraced and braced testing. An offset was evident across conditions and subjects. Since standing reference trials were standardised across conditions and subjects, the shift could be the result of the brace application or the different standing reference trials between conditions which created small deviations in alignment of the tibial and femoral anatomical coordinate systems. Therefore differences in movement patterns were reported rather than the absolute positions, i.e., the range from touchdown to maximum flexion instead of the (absolute) maximum flexion value. The following text describes average motion patterns for each subject and lists standard deviation values in parenthesis.

Ground reaction forces

Table 1 depicts mean peak vertical force (Fy) and mean peak anteroposterior shear force (Fx) calculated from each subject's respective unbraced and braced trials. Additionally, Table 1 identifies when foot-strike and peak Fy occurred during the landing-stance cycle. Subjects 1 and 3 exhibited similar peak vertical and peak posterior shear forces at foot-strike between bracing conditions. Subject 4 generated larger peak vertical force magnitudes with the unbraced knee than when supported. During unbraced testing, the data recording system failed to store ground reaction force data for subject 6. Consequently, only angular tibiofemoral data was used to determine whether jumping styles were similar between conditions. Mean peak vertical forces as shown in the Table 1 ranged from 2.161 (0.266) to 3.409 (0.358) and 2.369 (0.079) to 2.638 (0.592) for the unsupported and braced limb respectively. Mean peak posterior shear forces ranged from -0.637 (0.159) to -1.252 (0.174) during non-bracing and -0.603 (0.069) to 1.109 (0.111) when supported.

Table 1: Ground reaction force data Mean peak vertical and peak posterior ground reaction force normalised to body weight and mass of the brace across subjects and conditions

		Peak vertical force (Fy)		Peak posterior shear force (Fx)	
		Unbraced	Braced	Unbraced	Braced
Subject	Trials	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)
1	n = 5	2.947 (0.449)	2.612 (0.149)	-1.252 (0.174)	-1.109 (0.111)
3	n = 3	2.161 (0.266)	2.369 (0.079)	-0.637 (0.159)	-0.923 (0.090)
4	n = 5	3.409 (0.358)	2.638 (0.592)	-0.668 (0.067)	-0.603 (0.069)
6	n = 5	n/a	2.851 (0.301)	n/a	-1.102 (0.001)

Angular rotations

Flexion

As seen in Figure 5, tibiofemoral flexion curves were similar in shape between unbraced (solid bold line) and braced conditions (solid dashed line) and across subjects although differences in magnitudes were noted. Subjects 1 and 3 exhibited greater mean flexion ROM from touchdown through to peak flexion when the knee was braced. Conversely, a larger mean flexion ROM was evident in the unsupported knee for subjects 4 and 6 (Table 2). Following peak flexion, subjects 1, 4, and 6 stabilised the tibiofemoral joint and remained flexed overall whereas subject 3 came back into full extension.

Ab/adduction

Ab/adduction patterns were similar between bracing conditions but varied across subjects. For subjects 1 and 6, the lower limb was slightly adducted (foot brought in towards the midline of the body) until about peak Fy which placed the knee in relative varus. Conversely, subjects 3 and 4 abducted the tibia immediately at HS putting the knee into valgus position. Thereafter, all subjects abducted the tibia until about peak flexion forcing the knee into a valgus position. After stabilising the knee following peak flexion, the knee was positioned in relative varus. Additionally, subject 1 and 3 demonstrated relatively small ab/adduction movements about the neutrally positioned knee although subject 1 demonstrated greater abduction magnitudes when the knee was braced. Ab/adduction patterns for subjects 4 and 6 were offset from the baseline with greater magnitudes observed. Abduction ROM between conditions and across subjects are listed in Table 2. When the knee was supported, mean abduction ROM was reduced 1° and 3° for subjects 4 and 6 respectively. Conversely, abduction amplitudes increased 3° and 1° for subjects 1 and 3 during bracing.

Internal/external rotation

Internal/external rotational patterns were fairly similar between conditions and across subjects (Figure 4) with consistent ROM magnitudes except for subject 6 (Table 2). Subjects 3 and 6 exhibited little external knee rotation from HS to approximately peak Fy. All subjects demonstrated a *pronounced* internal rotation until peak flexion and remained internally rotated.

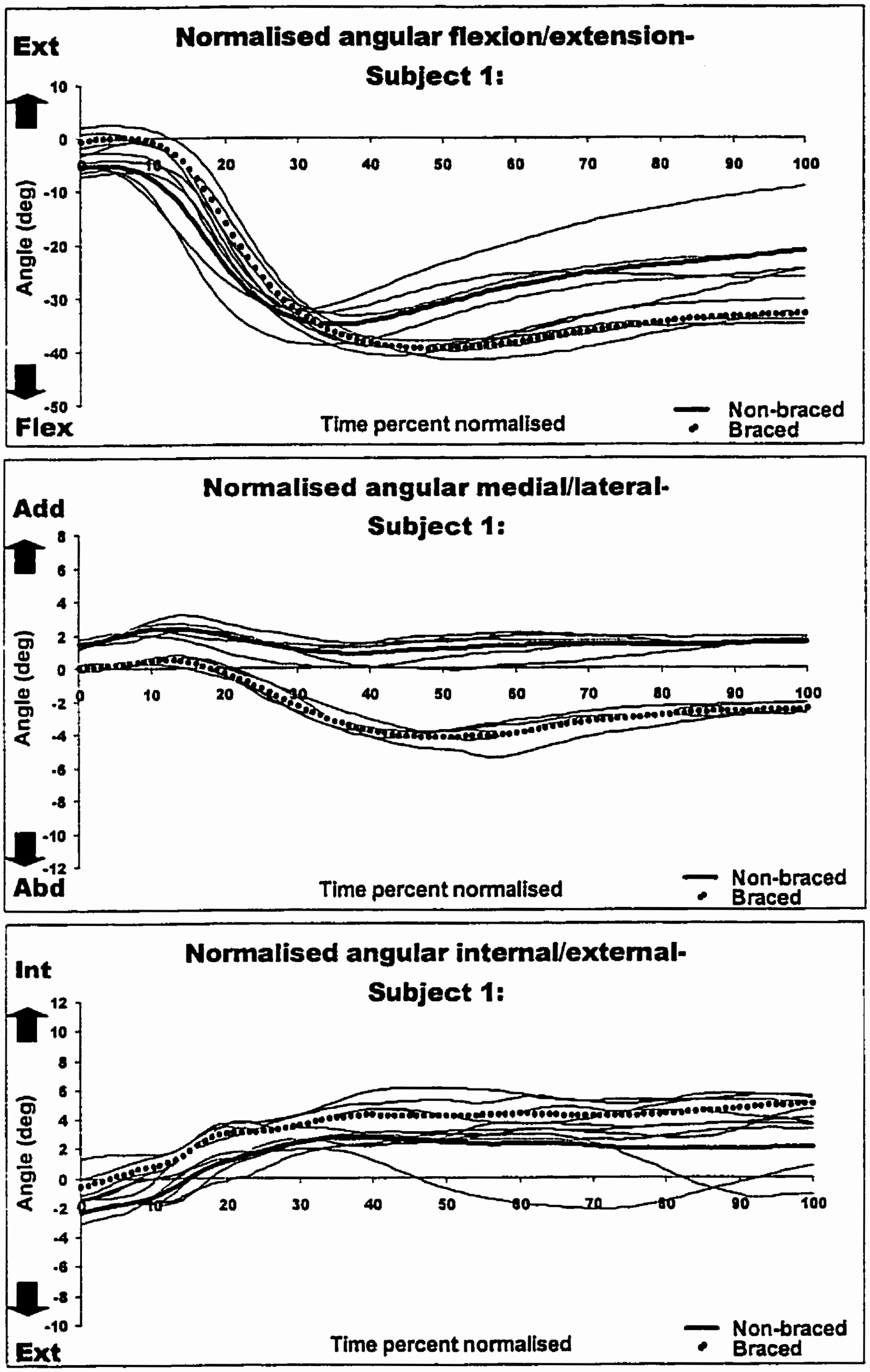

Figure 5 (a): Angular patterns of tibiofemoral joint motion derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

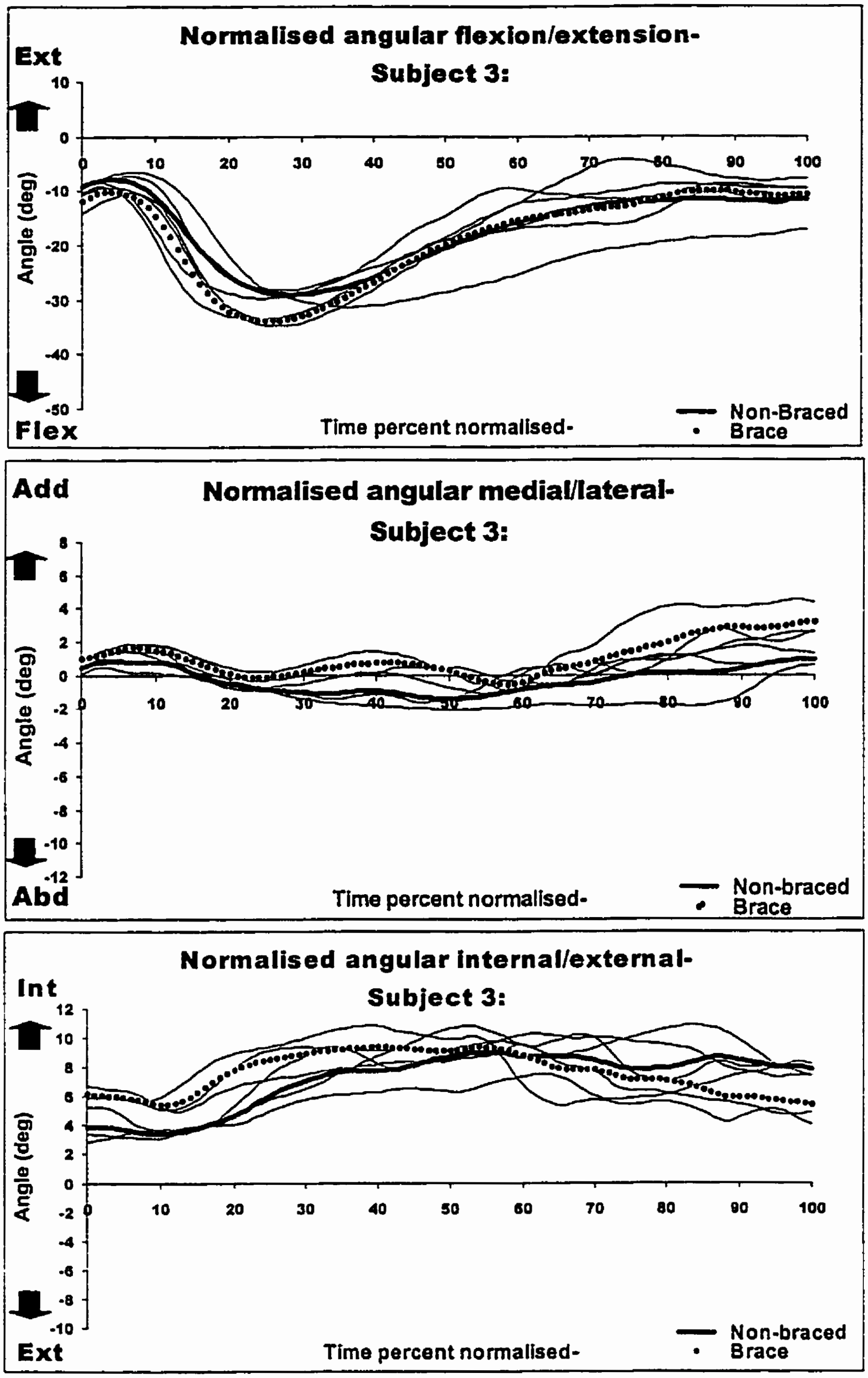

Figure 5 (b): Angular patterns of tibiofemoral joint motion derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

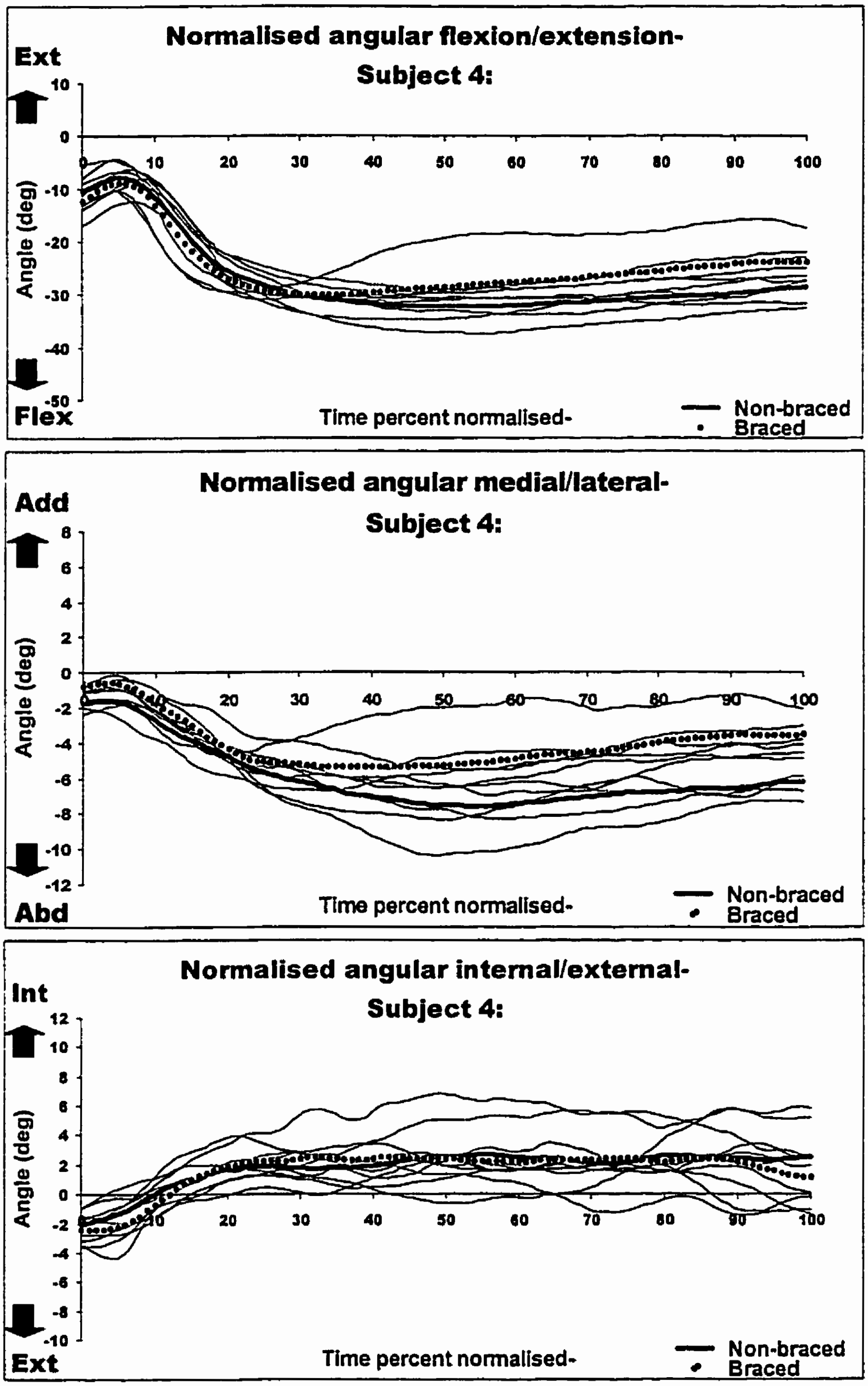

Figure 5 (c): Angular patterns of tibiofemoral joint motion derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

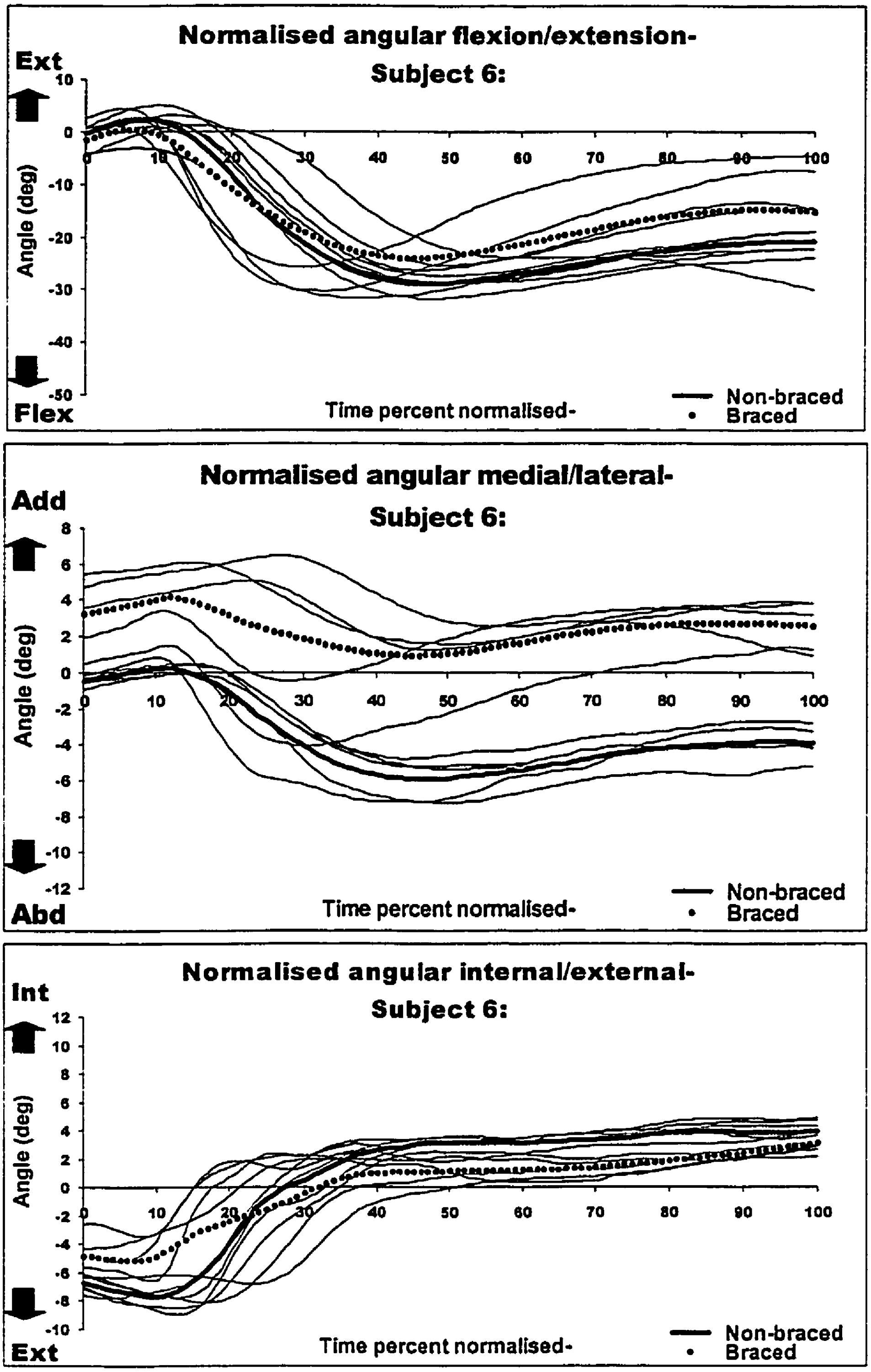

Figure 5 (d): Angular patterns of tibiofemoral joint motion derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

Table 2: Means of angular ranges of motion

		Flexion		Abduction		Internal Rotation	
Subject	**Trials**	**Unbraced**	**Braced**	**Unbraced**	**Braced**	**Unbraced**	**Braced**
1	n = 5	-29.9	-39.9	-1.5	-4.7	4.8	3.6
3	n = 3	-21.1	-23.7	-1.4	-2.1	5.3	4.4
4	n = 5	-24.2	-21.3	-5.9	-4.8	4.0	4.8
6	n = 5	-31.5	-24.6	-6.2	-3.2	10.8	5.8

Units in degrees

i) A negative value indicates that flexion of the TFJ took place.
ii) A negative value indicates TFJ abduction.
iii) A negative value indicates external rotation of the TFJ.

Bracing the knee reduced internal rotation magnitudes by 1°, 2° and 6° for subjects 1, 3 and 6 respectively but no changes were evident for subject 4 (Table 2).

Translations

Subjects average tibiofemoral joint translations for unbraced (solid bold line) and braced conditions (solid dashed line) are depicted in Figure 6.

Medial/lateral shift

With respect to the origins of the anatomical tibial and femoral reference frames, the least amount of movement excursions was mediolateral shift (Figure 6). Average shift patterns were similar in shape between bracing conditions although magnitudes varied considerably. Entirely different movement patterns were observed across subjects. Subject 1 demonstrated an initial lateral tibial shift from HS until about peak Fy averaging 2.9 mm and 2.7 mm across unbraced and braced conditions respectively. Thereafter until peak flexion, the tibia moved 1.2 mm medially when unsupported and remained constant thereabout. Bracing resulted in a larger medial tibial shift of 2.8 mm during flexion followed by a 1.3 mm lateral excursion when extending. Subject 3 and 4 exhibited little or no shift movements following foot-strike. At approximately 40% into the cycle, subject 3 experienced a medial tibial displacement. Magnitudes remained unchanged between the unbraced and braced conditions as medial movements amounted to 3.5 mm and 3.1 mm respectively. Similarly, lateral magnitudes were unaffected as the tibia moved 3.4 mm and 2.7 mm medially upon completion of the cycle. When the knee was unsupported, subject 4 demonstrated a small 1.0 mm lateral excursion until approximately peak Fy but none during bracing. Magnitudes remained unchanged between the unbraced and braced conditions. Medial movements amounted to 2.2 mm and 2.5 mm respectively until about peak flexion with lateral movements of 2.2 mm and 2.7 mm upon completion of the cycle. Since subject 6 exhibited the largest variability, unbraced and braced patterns only fairly agreed. Small lateral excursions (< 2 mm) were observed at HS during both brace conditions. When unsupported, a 5.1 mm medial tibial excursion was evident until about peak flexion. The tibia then moved laterally 1.5 mm before finally moving 3.00 mm medially towards the end of the cycle. During bracing, the tibia shifted laterally 1.7 mm until about peak Fy. Thereafter it shifted medially approximately 2.1 mm and remained thereabout until the of the cycle.

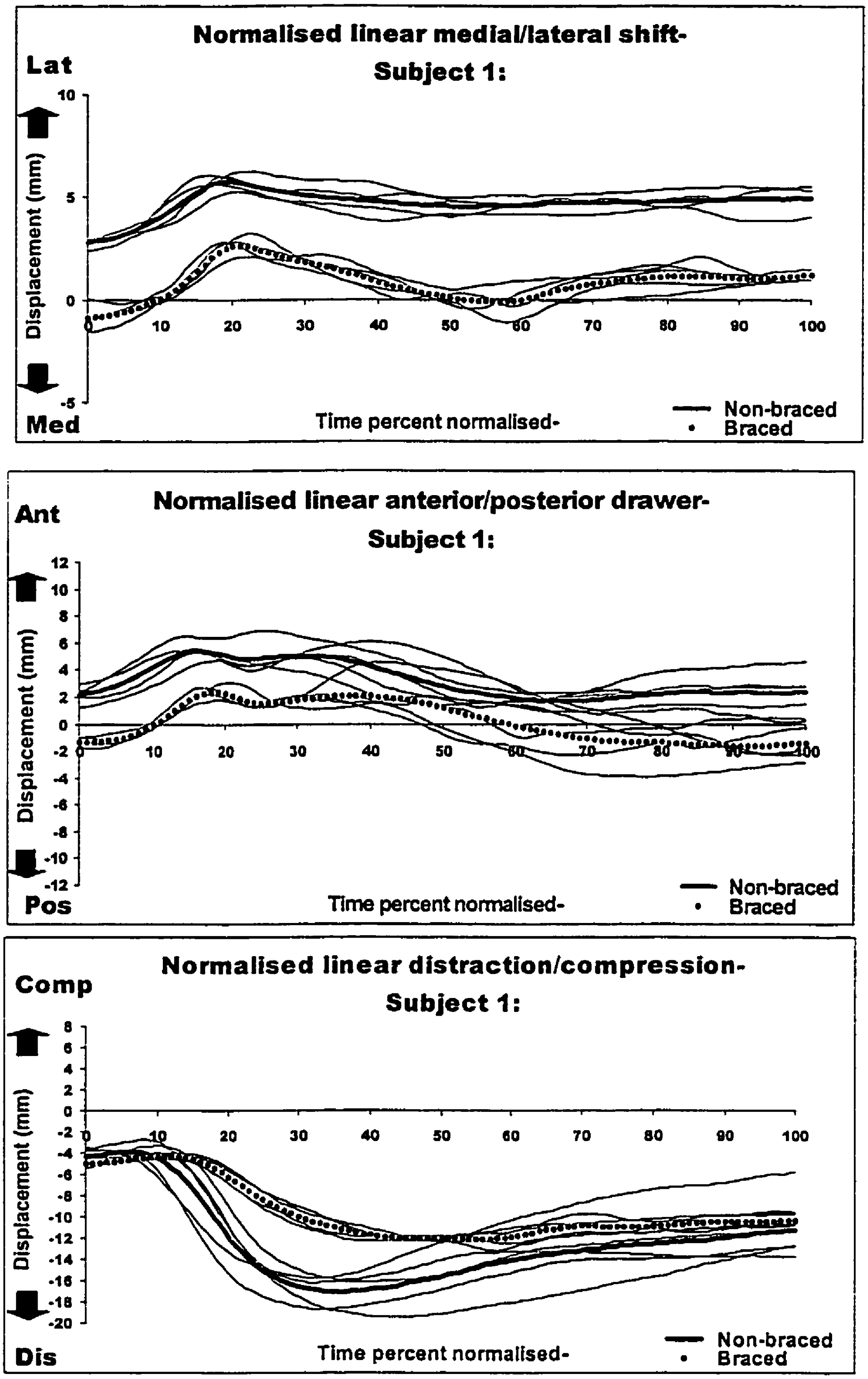

Figure 6 (a): Linear patterns of tibiofemoral joint translations derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

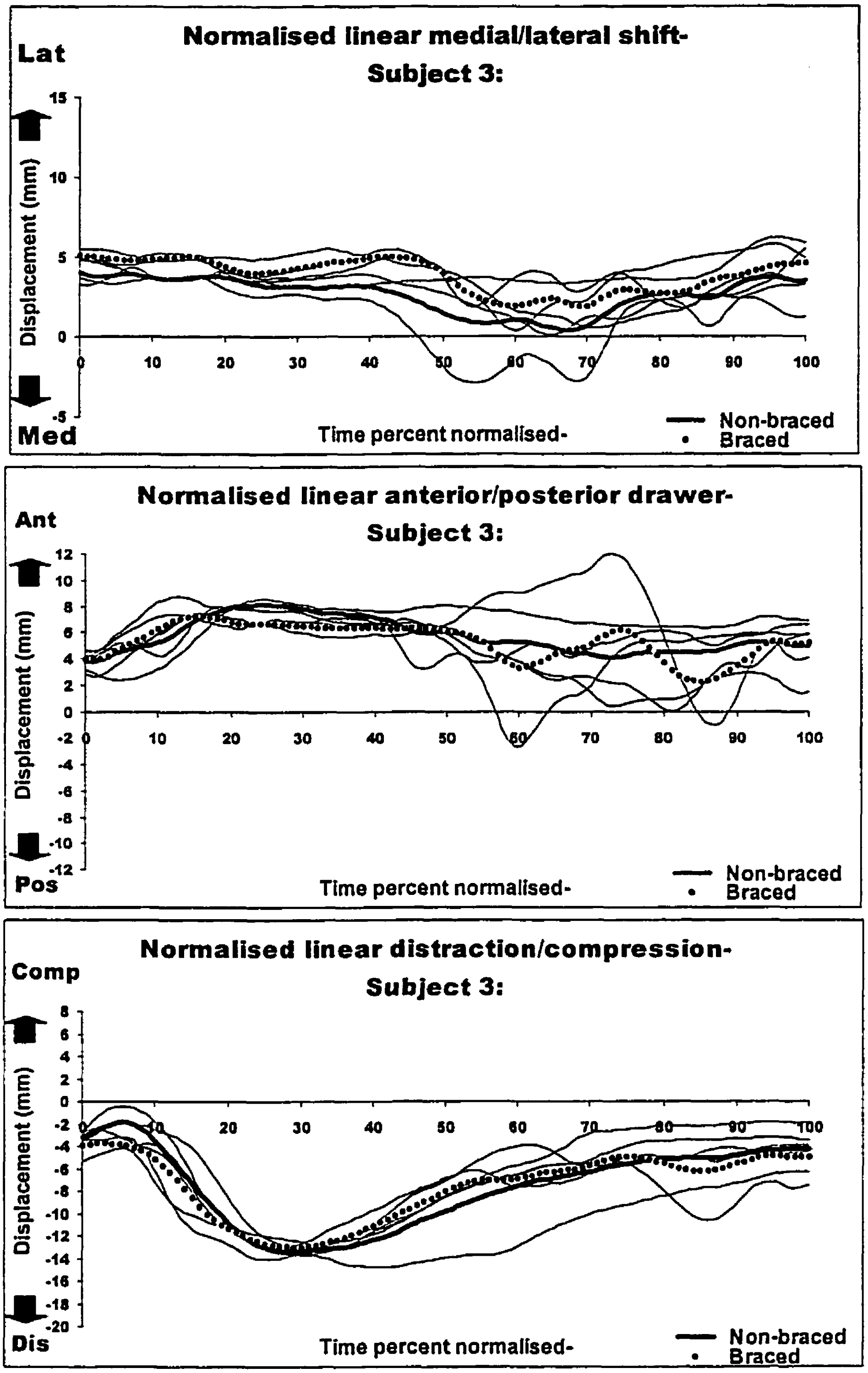

Figure 6 (b): Linear patterns of tibiofemoral joint translations derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

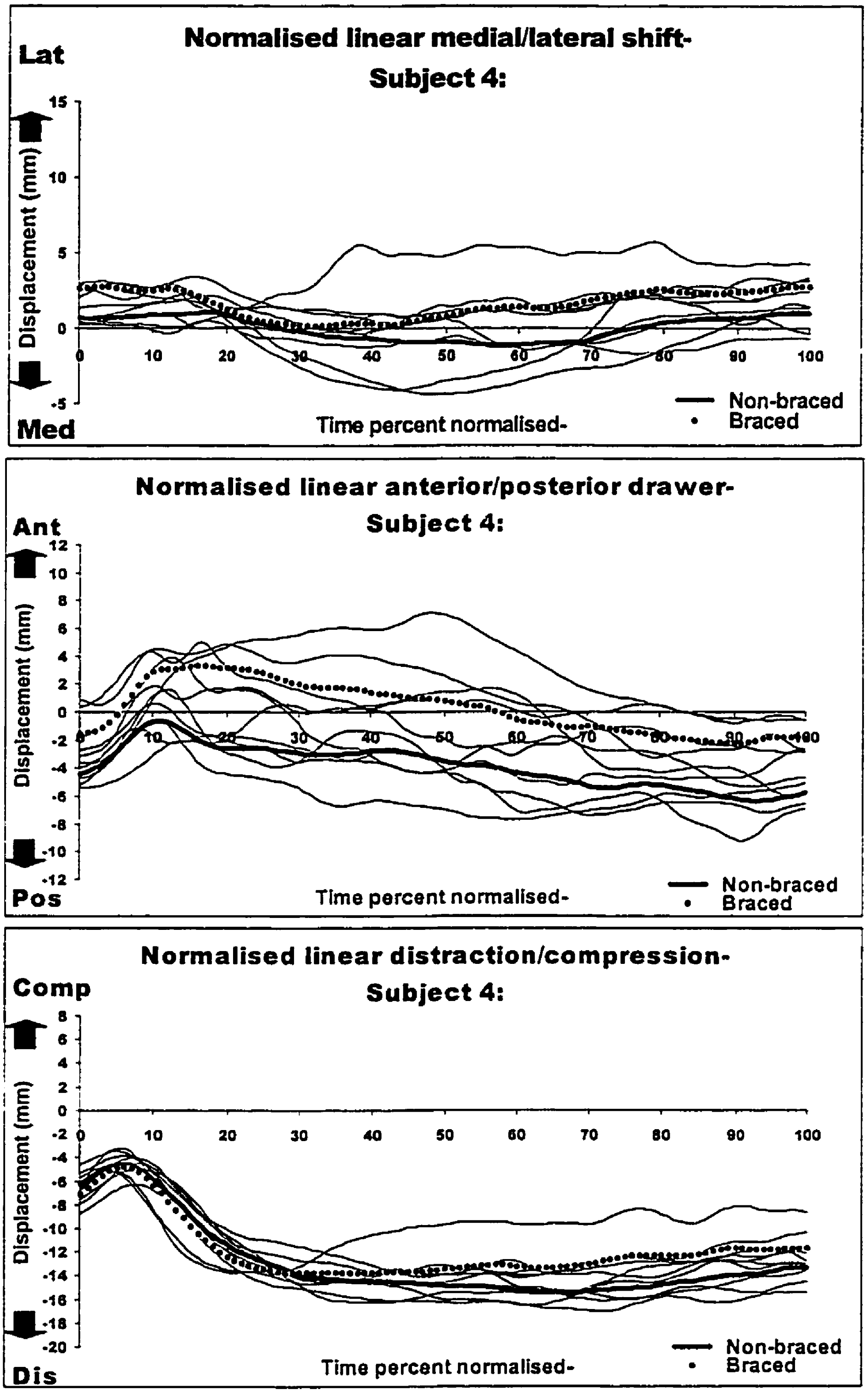

Figure 6 (c): Linear patterns of tibiofemoral joint translations derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

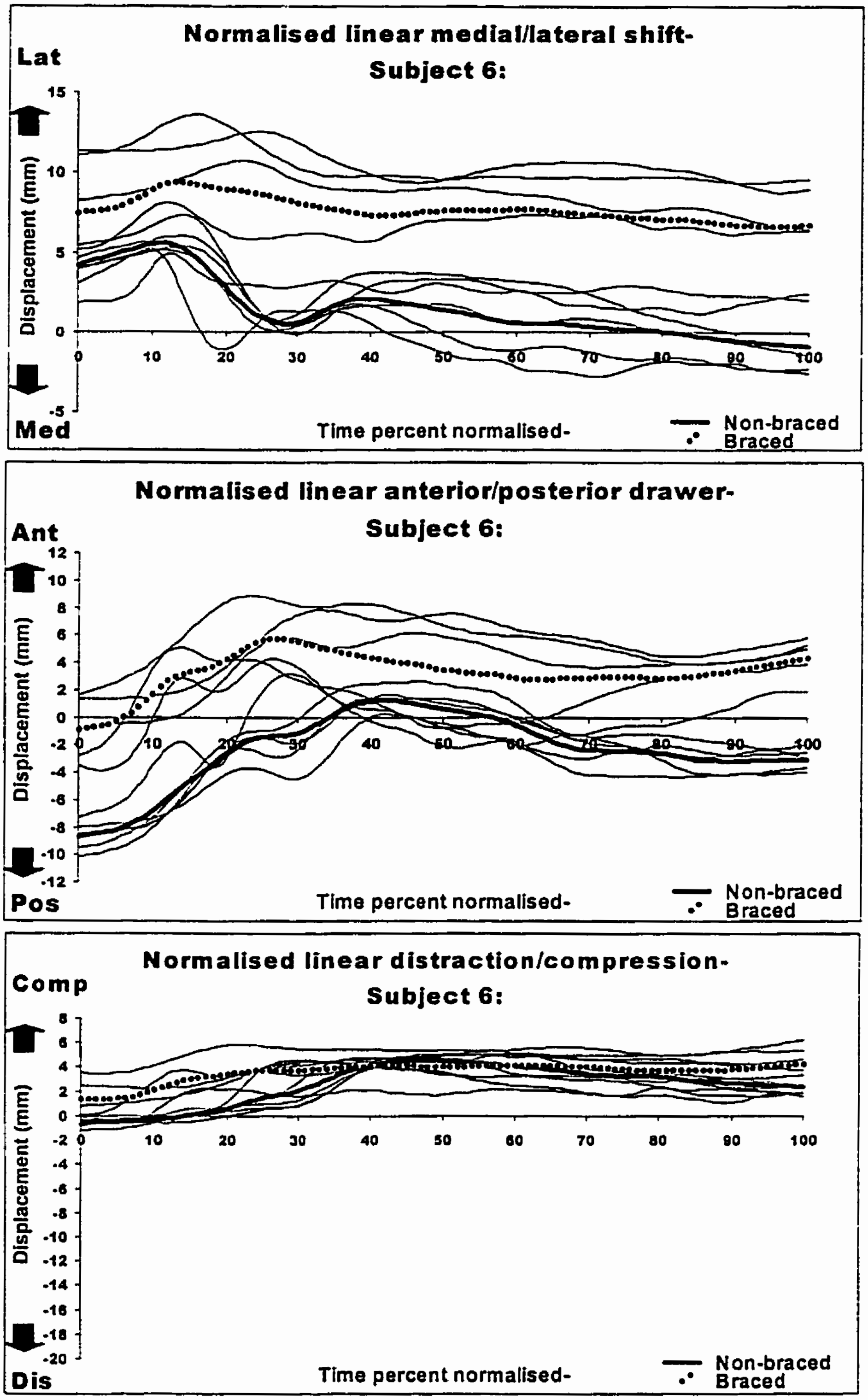

Figure 6 (d): Linear patterns of tibiofemoral joint translations derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics. (a) Subject 1; (b) Subject 3; (c) Subject 4; (d) Subject 6.

Table 3: Means of linear ranges of motion

		Medial shift		Anterior drawer		Distraction	
Subject	**Trials**	**Unbraced**	**Braced**	**Unbraced**	**Braced**	**Unbraced**	**Braced**
1	n = 5	-1.2	-2.8	3.0	2.7	-12.9	-7.8
3	n = 3	-3.5	-3.1	3.5	2.4	-11.5	-9.2
4	n = 5	-2.2	-2.5	2.2	3.5	-10.8	-8.9
6	n = 5	-5.1	-2.1	8.8	5.7	5.0	2.6

Units in mm

i) A negative value indicates the tibia remained in a medial position with respect to the femur even if it shifted laterally.
ii) A negative value indicates the tibia remained in a posterior position with respect to the femur even though it had moved in its most anteriorly located position.
iii) A negative value indicates that the joint was still compressed even though it was in its most distracted position.

Anterior/posterior drawer

Anterior/posterior drawer is described along the floating axis. As seen in Figure 6, anteroposterior drawer curves were similar in shape between bracing conditions and fairly similar across subjects although differences in magnitudes were noted. Overall, the tibia exhibited a rapid anterior displacement with respect to the femur from HS to approximately peak Fy. Thereafter the origin of the tibial reference frame was drawn posteriorly during flexion.

Anterior displacements remained unchanged for subject 1 (Table 3). Anterior excursions amounted to 3.0 mm and 2.7 mm for the unbraced and braced conditions with associated posterior movements of 3.7 mm and 4.0 mm, respectively. Conversely, two subjects demonstrated small reductions in anterior displacements during bracing. For subject 3, the unsupported tibia moved anteriorly 3.5 mm and 2.4 mm when braced. During flexion, the tibia moved 3.0 mm posteriorly and 4.0 mm when supported. Similarly, bracing reduced anterior drawer magnitudes from 8.8 mm to 5.7 mm for subject 6. Posterior movements were also reduced from 4.3 mm to 2.8 mm. The opposite was evident for subject 4 as anterior displacements were larger with the braced knee. Anterior tibial drawer amounted to 2.2 mm when the knee was unsupported and 3.5 mm when braced. Posterior displacements remained unchanged at 5.7 mm and 5.6 mm respectively.

Distraction/compression

Distraction/compression refers to the origins of the two anatomical coordinate systems being distracted or shortened and not to the contact or separation of the articular surfaces. The selected origins of the tibia and femur become farther apart with knee flexion, the result of the curvature of the femoral condyles.

As shown in Figure 6, distraction-compression curves were similar in shape across bracing conditions. Additionally, distraction-compression patterns for subjects 1, 3, and 4 exhibited a striking similarity with knee flexion-extension but subject 6 exhibited an entirely different pattern. As the knee extended prior to HS, subjects 1, 3 and 4 demonstrated little or no compression (Table 3). Distraction occurred from HS until peak flexion followed by compression accompanying knee extension. Interestingly, bracing reduced distraction magnitudes despite knee flexion magnitudes being larger for subjects 1 and 3. Subject 1 demonstrated average unbraced and braced joint distractions of 12.9 mm and 7.8 mm with associated compressions of 5.8 mm and 1.8 mm, respectively. The lower compression magnitudes during bracing may be attributed to the subject remaining more in a flexed position. Similar magnitudes were observed for subject 3 and 4. Unsupported distractions for subject 3 amounted to 11.5 mm and 9.2 mm when braced with associated compressions of 9.2 mm and 8.1 mm respectively. Subject 4 demonstrated distractions of 10.8 mm and 8.9 mm. The slight knee extensions following peak flexion resulted in small joint complex compressions of 2.1 mm for both conditions. Conversely, subject 6 exhibited compressions during flexion and distraction accompanying knee extension with reductions of distraction-compression magnitudes during bracing. The joint complex underwent compression of 5.0 mm from HS to about peak flexion and 2.6 mm for the unbraced and braced conditions respectively. Subject 6 exhibited a distraction of 2.2 mm when unsupported with no distractions observed during bracing.

Discussion

Recent investigations have implemented invasive markers to directly measure tibiofemoral joint motion during walking and light running [8,9,10,11]. Standard deviations less than 0.6° for rotations and translations less than 0.4 mm have been reported when comparing RSA values and MacReflex data recorded in a volume of 0.25 m^3 [25]. In those studies, knee joints were clinically evaluated to be "within normal limits" with no pathologies. The procedure is justified since the accuracy of skin markers and exoskeletal linkage systems is questionable. For this investigation, six subjects with partial or complete ACL rupture had intracortical pins implanted into the tibia and femur to examine the relative tibiofemoral joint motion between unbraced and braced conditions during strenuous activity.

Average peak vertical force at foot-strike and the peak anterior-posterior shear force were generally consistent between unsupported and braced conditions. However magnitudes varied across subjects owing to the fact that subjects jumped within their own comfort limits. The consistency between brace conditions indicate jumps onto the force platform were similar. Therefore, changes in skeletal kinematics cannot be attributed to differences in jumps onto the force platform but rather to the brace itself.

Tibiofemoral rotations and translations show a general trend across subjects, i.e. the shape and amplitudes of the skeletal marker based curves were similar. The major difference is a shift between the unbraced and braced trials. This can be attributed to the brace or the different standing reference trials used during the unbraced and braced trials creating small differences in alignment of the tibial and femoral anatomical coordinate systems [22]. Generally, intra-subject differences between unbraced and braced patterns were small, i.e. knee kinematics were very repeatable. Inter-subject differences were typically much larger than intra-subject variability. Differences mainly consisted of amplitudes and positional changes at touchdown.

Angular rotations

Flexion

As expected, tibiofemoral flexion was the largest component of total knee motion. All subjects demonstrated fairly similar flexion patterns between unbraced and braced conditions although flexion ranges of motion varied. Subject 1 and 3 exhibited greater flexion magnitudes (30° vs. 40°) and (21° vs. 24°) respectively when the knee was supported. Conversely, subject 4 and 6 demonstrated larger ROM when unsupported (24° vs. 21° and 32° vs. 25°).

Flexion patterns corresponded well although flexion magnitudes were greater compared to walking investigations by Lafortune et al. [8] and Reinschmidt et al. [11,22]. Lafortune et al. [8] and Reinschmidt report the knee was slightly flexed between 0-10° at heelstrike and continually increased to about 15°-20° in stance. Total range of motion of the tibiofemoral joint was approximately 20° - 30° degrees [11,22,8]. Additionally, flexion/extension patterns were similar in shape and amplitude with respect to McClay [9] and Reinschmidt et al. [11,22] investigations which assessed tibiofemoral motion during running. McClay [9] reported flexion ranges of motion of 21° for the normal group and 26° for the patellofemoral pain group whereas Reinschmidt [11,22] found larger ROM magnitudes. The discrepancies across investigations are likely the result of different definitions of the tibial and femoral anatomical coordinate systems. McClay [9] and Lafortune et al [8] employed roentgen-stereo-photogrammetric analysis (RSA) whereas Reinschmidt utilised the neutral standing trial and RSA methods.

Ab/adduction

The patterns and magnitudes of ab/adduction during the landing-stance phase varied across subjects. Subjects 1 and 6 demonstrated small lower limb adduction magnitudes of less than 1.0° at HS followed by knee abduction during flexion. Subjects 3 and 4 abducted immediately following HS but subject 3 demonstrated smaller ab/adductions when the knee was supported. Subjects 1, 3, and 4 demonstrated greater mean adduction ROM from touchdown through to peak flexion when the knee was braced. Conversely, a larger mean adduction ROM was evident in the unsupported knee for subject.

The ab/adduction patterns and magnitudes from this investigation were in total disagreement with previous skeletal tibiofemoral investigations. Lafortune et al. [8] reported that no tibiofemoral ab/adduction movements took place during the stance phase of walking [8]. Reinschmidt et al. [11] found patterns varied across subjects with ROM as high 10°. In running studies, McClay [9] found all subjects exhibited similar skeletal tibiofemoral ab/adduction patterns. Initial adduction motion averaged 6° from touch down followed by a gradual abduction until the end of stance. Interestingly, larger adduction amplitudes were observed for the injured group. Conversely, Reinschmidt et al. [10,22] reported *skeletal* ab/adduction patterns and magnitudes varied considerably between subjects. Two subjects had initial abduction movements of 6° and 9° until midstance followed by an adduction until the end of stance. One subject exhibited a small adduction movement of 4° and a 3° adduction towards the end of stance. The abduction and adduction patterns that occurred during flexion/extension from this investigation were in general agreement with Reinschmidt et al. [22] although magnitudes were lower.

The difference in results across studies and the finding of this study may be explained by differences in defining the anatomical coordinate systems of the femur and tibia. Lafortune et al.

[8] and McClay [9] employed anatomical coordinate systems based on a roentgen-stereophotogrammetric analysis whereas Reinschmidt [10,11,22] utilised the neutral standing trial. Also, it would be expected patterns and magnitudes may vary dependent on the activity involved.

Ab/adduction ranges of motion is limited to approximately 5° due to ligamentous restriction and geometry of the knee [26]. Reinschmidt [22] speculated that ab/adduction ROM may even be smaller during high dynamic activity since the knee is loaded and stabilised by muscular forces. The large adductory magnitudes found for subjects 1, 4 and 6 may not reflect "true" ab/adduction patterns. Reinschmidt [22] reported small non-primary rotations are highly susceptible to cross-talk from flexion-extension. This stems from alignment problems of the anatomical coordinate systems the result of which movements may exceed and mask the actual motion. Ab/adduction patterns for subjects 1, 3 and 4 were very similar to the flexion-extension motion, giving rise to speculations that the relatively large abductions were mainly caused by cross-talk.

Internal-external rotations

The internal-external rotations from this investigation compare favourably with patterns and magnitudes from other investigations [8,9,10,11,22]. In the Lafortune et al.[8] walking study, all subjects exhibited initial internal knee (tibial) rotations ranging from 2° to 6° with respect to the femur. Reinschmidt [11] reported greater intersubject variability. Two subjects demonstrated internal rotations of 5° and less than 2° at HS while the other subject externally rotated 4° at HS. McClay (1990) found similar internal/external patterns but with larger magnitudes. Slight differences in magnitudes were evident between the normal and the pathological group respectively. The knee internally rotated (10.7° vs. 12.4°) from touchdown to midstance followed by an external rotations (11.7° vs. 11.8°) across groups. Reinschmidt (1996) reported that motion was highly subject dependent. Based on neutral standing trials, skeletal internal/external patterns were fairly similar across subjects although the magnitudes varied between subjects. From heelstrike to midstance, subjects demonstrated either a pronounced initial internal rotation varying between 7°- 9°or a small rotation of 2°. During the later half of stance, the knee externally rotated. Reinschmidt reported patterns and magnitudes compared favourably with McClay's (1990) investigation although in that study the magnitudes of internal rotation were higher than external rotation during the second half of stance. Additionally, Reinschmidt's skeletal internal/external curves based on both the neutral standing trial and RSA agreed well but an obvious offset between the two average curves was evident. Furthermore, Reinschmidt found noticeable differences for internal knee rotations between the two coordinate systems not evident for the other rotations. Initial internal knee rotations based on RSA and neutral standing trials averaged 11° and 7° respectively. Small deviations and inconsistencies in defining the anatomical coordinate systems may account for the differences across subjects.

Translations

Tibiofemoral translations were in general agreement with Reinschmidt [22] study. Lafortune et al. [8] and McClay [9] associated flexion with medial shift, posterior drawer and tibial distraction with the opposite being true for extension. Conversely, Reinschmidt [22] related flexion with lateral shift, posterior drawer and tibial distraction while extension demonstrated contrary patterns. Although magnitudes and patterns between investigations varied, the differences can be attributed to differences in locomotor activity and to differences in the placement of the anatomical axes. Lafortune et al. [8] and McClay [9] described linear displacements as absolute values relative to the positions of the tibia and femur at heel strike. Reinschmidt [22] reported translations as changes in movement between the origin of the femoral and tibial anatomical coordinate systems already some distance apart.

Medial/lateral shift

The least amount of translationary movement was mediolateral shift during stance. The patterns of mediolateral shift were similar to patterns reported by Reinschmidt [22] but generally in the opposite direction to what has been reported by Lafortune et al. [8] and McClay[9] . Lafortune [8] reported tibial shifts closely matched the patterns of knee flexion/extension. When the knee flexed early during stance, the tibia shifted 2.3 mm medially initially followed by a lateral 1.5 mm shift as the knee extended during the middle part of stance [8,20]. All subjects in McClay's study exhibited a medial translation during the first half of stance followed by a lateral translation until toe-off. Reinschmidt [22] reported no consistent patterns during the first 15% of stance afterwhich the tibia underwent a lateral shift with respect to the origins of the tibial and femoral reference frames. From 40% to 80% stance, the tibia shifted medially followed by a lateral shift during the last 20% of stance. Reinschmidt's [22] RSA based mediolateral patterns were generally in the opposite direction from McClay [9]. reported. This may be related to the discrepancies found for the ab/adduction motion.

Anterior/posterior drawer

This investigation is the first to directly measure a skeletal anterior tibial displacement when performing a dynamic functional task prior to the tibia being drawn posteriorly during flexion. Overall, the tibia exhibited a rapid anterior displacement with respect to the femur from HS to approximately peak Fy. Thereafter the origin of the tibial reference frame was drawn posteriorly during flexion. Previous investigators have reported a posterior tibial displacement during flexion with the converse being true for extension.

During normal walking, Lafortune et al. [8] reported a posterior tibial displacement of 3.6 mm during flexion and an anterior excursion of 1.3 mm past the neutral position (defined as 0 mm) during extension. McClay [9] found similar anterior/posterior patterns during running although magnitudes were larger. At heelstrike, the origin of the tibial anatomical reference frame was positioned posterior to the femoral origin and continued moving posteriorly during the first 25-35% of the stance cycle as the knee flexed. Thereafter the tibia moved anteriorly during extension. Posterior excursions amounted to 3.9 mm and 2.8 mm for the normal group and patellofemoral pain group respectively. Throughout stance, the tibial reference point was more anterior for the pathological group and they demonstrated greater anterior drawer 2.4 mm and 7.3 mm respectively. Reinschmidt [22] observed negligible translations during the first 5% of stance, which was followed by a posterior tibial displacement of 4 mm between the origins of the tibial and femoral reference frames. From about 40% to 80% of stance, the tibia moved anteriorly (5mm) followed by a fast posterior displacement towards the end of stance.

Distraction/compression

Although the articular surfaces come together during loading as the knee flexes, the selected origins distract as a result of the curvature of the femoral condyles during flexion [9]. Lafortune et al. [8], McCaly [9] and Reinschmidt [22] all report similar tibiofemoral distraction-compression patterns with a striking similarity with knee flexion and extension. After heelstrike, Lafortune et al. [8] found the joint complex distracted 3.2 mm during flexion followed by a 0.2

mm compression accompanying knee extension. McClay [9] reported a distraction of 4.4 mm for the two normal subjects and 5.9 mm for the two subjects with patellofemoral pain syndrome. Compression magnitudes during extension 4.4 mm and 3.3 mm. Reinschmidt [22] observed a 5.6 mm distraction between the origins of the two coordinate systems from 10% to 40% of stance followed by an even larger compression of 6.8 mm. During the final 20% of stance, a distraction movement of 2.8 mm was observed.

Cardan angles and joint translations calculated using the joint coordinate system (JCS) is highly susceptible to alignment errors and uncertainties in defining the anatomical coordinate system [22]. Small deviations in alignment of the anatomical frames of reference across subjects make inter-subject comparisons difficult. This is a concern since differences may be partially (or totally) caused by slight differences in defining the anatomical coordinate systems.

Reinschmidt [22] has indicated that translations may be dependent on the rotations. Because the orientation and location of the anatomical tibial and femoral reference frame are based on anatomical landmarks, the points can be considered "arbitrary", meaning they are likely not to reflect an "average" joint centre. If the origins of the two coordinate systems do not coincide with an average joint centre or if such an average does not exist, the translations are very much dependent on the rotations. Cross-talk would register a translation even though a pure rotation would take place. Blankevoort et al. [27] suggested meaningful distances be calculated between points embedded in the two bodies (e.g. ligament insertion sites) which would provide more comprehensive and physiological meaningful translations than translations calculated along the axis of a joint coordinate system.

Conclusion

Intra-subject peak vertical forces and anterior shear forces were generally consistent between unsupported and braced conditions indicating jumps onto the force platform were similar. The small intra-subject angular and translational differences across conditions cannot be attributed to variations in jumping styles, but rather to the brace itself. Tibiofemoral rotations and translations show a general trend across subjects, i.e. the shape and amplitudes of the skeletal marker based curves were similar. The major difference was a shift between the unbraced and braced trials. The offset between conditions can be attributed to the brace or the different standing reference trials used during the unbraced and braced trials. This created small differences in alignment of the tibial and femoral anatomical coordinate systems rather than to application of the brace itself. Generally, intra-subject knee kinematics were very repeatable but differences between unbraced and braced patterns were small. This may be due to the invasiveness of this protocol, that landings are performed onto a deficient limb, and that subjects jumped within their own comfort limits which did not maximally stress the ACL. As expected, inter-subject differences were typically much larger than intra-subject variability. Differences mainly consisted in amplitudes and position at touchdown.

Although this study included a small number of subjects, the information regarding bracing the ACL deficient knee and its effect on three-dimensional tibiofemoral joint motion *in-vivo* has been valuable. However, by increasing the number of subjects it would be possible to perform inferential statistical analyses. With continued tibiofemoral research *in-vivo*, the restraining effects of functional knee braces during strenuous activity would be better understood.

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Appendix A

Background

The purpose of this investigation is to evaluate whether Functional Knee Braces reduce anterior translational and rotational displacements for anterior cruciate deficient (ACLD) knees. With the proliferation of new functional braces claiming to stabilise ligament deficient knees, clinical and laboratory research is necessary to substantiate their effectiveness (Branch *et al.*, 1989; Cook *et al.*, 1989; Branch and Hunter, 1990). Early studies reported bracing appears to be effective during controlled low load static manoeuvres (Branch *et al.*, 1989; Cook *et al.*, 1989; Branch and Hunter, 1990; Vailas *et al.*, 1990; Beynnon *et al.*, 1992; DeVita *et al.*, 1992; Vailas and Pink, 1993). However, braces failed when high loads were encountered or when the load was applied in an unpredictable manner (Branch *et al.*, 1989; Cook *et al.*, 1989; Branch and Hunter, 1990).

Quantitative kinematic analysis is an important tool for gaining a thorough understanding of normal and pathological joint function during human locomotion (Reinschmidt, 1996; Reinschmidt *et al.*, 1997b). By developing normal joint profiles, identifying abnormalities is possible. This helps to improve diagnosis and treatment, the design and performance of reconstructive surgery, rehabilitation programs, the development of accurate biomechanical models, and the development or modification of functional knee braces.

Both tibiofemoral and patellofemoral kinematics have been extensively investigated using reflective markers attached to the surrounding soft tissue of the calf and thigh. However, surface markers may not accurately represent the underlying bone motion during dynamic activity (Reinschmidt, 1996; Reinschmidt *et al.*, 1997b). The relative movements between skin and markers and the underlying bone may introduce large errors (Nigg and Cole, 1994). This is a particular concern during high dynamic activity. Consequently, knowledge about skeletal tibiofemoral kinematics is limited, in particular abduction-adduction, internal-external rotations, and associated 3D linear displacements. Considerable questions remain regarding motion of the knee. A way to avoid the problem of surface markers is to use intracortical pins affixed with markers to directly measure skeletal motion. Recent investigations have surgically implanted intracortical pins in order to directly measure three-dimensional skeletal tibiofemoral and patellofemoral joint motion (Lafortune, 1984; McClay, 1990; Lafortune *et al.*, 1992; Reinschmidt, 1996; Reinschmidt *et al.*, 1997b). However, the applicability of such methods is limited due to the invasiveness of such procedures and the methodological concerns associated with this procedure.

To date, little research has examined the efficacy of functional bracing on the osteokinematics and arthrokinetics during high physiologic conditions. Since braces are designed for athletic activity, they should be evaluated under these conditions.

Statement of the Problem

It has been reported functional knee braces are effective in reducing anterior translations when subjected to static or low anterior shear forces but not during high dynamic activity (Branch *et al.*, 1989; Cook *et al.*, 1989; Vailas *et al.*, 1990; Cawley *et al.*, 1991; DeVita *et al.*, 1992; Vailas and Pink, 1993). Braced knees continue to give way under dynamic conditions and fail in situations (Cook *et al.*, 1989; Vailas *et al.*, 1990; Cawley *et al.*, 1991; DeVita *et al.*, 1992; Vailas and Pink, 1993) where high loads are encountered or when the load is applied in an unpredictable manner (Branch *et al.*, 1989).

Currently, little research has been conducted to examine the shielding effects of functional braces on 3D osteokinematics and arthrokinetics during high physiologic conditions. Knowledge about skeletal translations and rotations are limited, in particular ab/adduction, internal/external rotations and 3D linear translations. The differences in the reporting of frontal and transverse motion may be attributed to variations in experimental designs and as such comparisons between investigations are nearly impossible (Vailas and Pink, 1993). For a complete kinematic analysis, three-dimensional motion analysis is required whereby all six degrees of freedom (three rotations and three translations) can be discerned (Branch *et al.*, 1989; Branch and Hunter, 1990).

Since relative movements between skin markers and underlying bone introduce large errors during high dynamic activity, invasive markers affixed directly to the tibia and femur provide the most accurate means for measuring bone movements (Cappozzo, 1991; Nigg and Cole, 1994). By directly recording 3D skeletal motion for a group of anterior cruciate deficient (ACLD) subjects, this investigation seeks to determine whether rotations and translations are reduced with the knee functionally braced during strenuous activity. Emphasis will to discern whether anterior translations are reduced during braced conditions when performing a dynamic *One Legged Hop* (OLH).

Hypothesis

Evidence suggests braced knees continue to give way during strenuous dynamic activity and fail in situations where high loads are encountered or when the load is applied in an unpredictable manner. The criteria for determining whether ACL reconstructive surgery is required is based on the patients' functional instability and their instability tests scores using physical and instrumented tests (e.g. KT1000). Therefore there is a need to quantify true anatomical tibiofemoral motion. This study will involve new techniques including intracortical pin implantation and 3D-motion analysis to assess 3D kinematics in an ACLD knee during a functional (OLH) manoeuvre.

The focus of this investigation is to determine whether application of a functional brace to an ACLD knee reduces abnormal tibial displacements (rotational and translational) during conditions of strenuous activity. In particular, during high dynamic loading, are kinematic differences evident (e.g. anterior translations) between functional knee brace application to conditions where no brace is worn?

Rationale

Knee stability arises primarily from two restraint systems; a *passive restraint system* which is comprised of the ligaments and capsule and a *dynamic restraint system* consisting of the neuromuscular elements (Wojtys and Hutson H.J., 1994). In an unloaded knee, all externally applied forces or moments are internally resisted by the ligaments and capsule. Whereby the primary role of the ACL is to resist anterior-posterior translation, functional knee braces are designed to reduce anterior-translational and rotational displacements for ACLD knees (Vailas and Pink, 1993). This provides functional stability to the unstable knee and subsequently enhances athletic performance by reducing pathological subluxation of the joint. However, when engaged in athletic activities, subjects continue to report episodes of knee instability despite wearing a brace. Previous manual knee evaluations including the *anterior drawer test*, the *pivot shift test*, and *Lachman's test* have all measured tibial displacements during simulated static loading conditions but they do not reflect true physiologic loading (Branch *et al.*, 1989).

For an improved understanding of the effects of functional bracing, this analysis is important in defining the parameters of 3D tibiofemoral motion so that in the future pathological motion due to ACL insufficiency can be better understood, diagnosed and treated effectively.

Limitations

Limitations within this investigation can be summarised as follows:

Skeletal kinematic recordings employing intracortical pins is an invasive procedure. This may cause discomfort and the anaesthetics may alter the subjects' perception. However, previous bone pin research has reported subjects did not experience significant discomfort. Subjects reported they moved their knees freely and their walking and running styles remained unaffected (Lafortune, 1984; McClay, 1990; Lafortune *et al.*, 1992; Reinschmidt, 1996; Reinschmidt *et al.*, 1997b).

To sufficiently stress the anterior cruciate ligament (ACL), subjects jump for maximal horizontal distance by pushing off from their sound limb and land using their deficient limb. Due to the invasiveness of this protocol, that jump landings are performed onto a deficient limb, and subjects jump within their own comfort limits, this may not be adequate to maximally stress the ACL to yield differences between test conditions.

The MacReflex calibration frame required to calibrate the measurement area is limited due to the insufficient number of calibration points (nine). The accuracy of spatial reconstruction is reduced when a small number of calibration points are used (Hatze, 1988).

Cardan angles derived using the Joint Co-ordinate System (JCS) (Grood and Suntay, 1983) may not be adequate in describing tibiofemoral joint motion (Reinschmidt, 1996). Non-primary rotations may be highly influenced by cross talk from tibiofemoral flexion-extension. Translations may be calculated based on more meaningful distances. For example, between two points embedded in both segments such as the insertion sites of ligaments. Such distances may provide more comprehensive and meaningful translations (Reinschmidt, 1996).

Review of Literature: Kinematics

From the following review of literature, the kinematic component has been published and can be referenced as follows: Ramsey, D.K. and Wretenberg, P.F. (1999) Biomechanics of the knee: Methodological considerations in the *in-vivo* kinematic analysis of the tibiofemoral and patellofemoral joint. *Clin Biomech* 14/9 595-611.

Literature review

Knee stability arises primarily from two restraint systems; a *passive restraint system* which is comprised of the ligaments and capsule and a *dynamic restraint system* consisting of the neuromuscular elements (Wojtys and Hutson H.J., 1994). Numerous biological, anatomical and biomechanical studies report that the anterior cruciate ligament (ACL) is the primary knee stabiliser which resists excessive anterior translation of the tibia relative to the femur (Lafortune, 1984; Lafortune *et al.*, 1992; Ahmed *et al.*, 1992; Smith *et al.*, 1993; Beynnon *et al.*, 1994). Currently, no methodologies exist in the literature regarding accurate measuring techniques for measuring anterior tibial translation (ATT). However previous investigations have utilised different external or in *vitro* techniques (Marans *et al.*, 1989; Ahmed *et al.*, 1992) . Evidence now supports the notion that chronic ACL deficiency results in significant knee instability (Smith *et al.*, 1993). Advances in computerised motion tracking systems now enable complete 3D segmental kinematic analysis and aid in describing knee motion. The improved understanding of knee stability and ACL function during dynamic activities may facilitate improvements in the design of rehabilitation programs, of reconstructive surgery, the development of accurate biomechanical models, and the development of functional knee braces etc (Beynnon *et al.*, 1992; Beynnon *et al.*, 1992).

This review primarily focuses on the use of intracortical pins to document three-dimensional tibiofemoral and patellofemoral kinematics during performance of a dynamic activity. Knowledge about skeletal translations and rotations are limited, in particular ab/adduction and internal/external rotations. Specific to *in-vivo* investigations, differences in defining of the anatomical co-ordinate system may account for the variations of non-primary rotations reported in the literature (McClay, 1990). Therefore, emphasis is on methodological concerns since ab/adduction and internal/external rotations are small and these non-primary rotations may be highly influenced by cross talk (from knee flexion/extension) derived from alignment problems of the anatomical co-ordinate systems (Reinschmidt, 1996; Reinschmidt *et al.,* 1997b).

Articulations of the Knee Joint

The anatomy of the knee joint has been well documented in the literature and will not be further described in this paper. While the movements of the knee are stabilised and guided by ligaments, their major functions are to attach articulating bones to one another across a joint, guide movements, maintain conformable joint fittings (congruency) and act as strain sensors for the joint (Frank and Shrive, 1994). In particular, the ACL is responsible for 85% of the total restraining force in preventing excessive anterior translation of the tibia relative to the femur and secondly limits varus-valgus and axial tibial rotations of the knee (Lafortune, 1984; Lafortune *et al.,* 1992; Smith *et al.,* 1993; Takeda *et al.,* 1994; Frank and Shrive, 1994). When ligamentous instability exists, these translational components become even larger (Marans *et al.,* 1989).

Articulations occur between the proximal end of the tibia and the posterior surface of the patella with the distal end of the femur being the common participant (McClay, 1990). These consist of three translations; anterior-posterior drawer, medial-lateral shift, distraction-compression, and three rotations; internal-external, abduction-adduction and flexion-extension (McClay, 1990; Lafortune *et al.,* 1992; Reinschmidt, 1996; Reinschmidt *et al.,* 1997b). Figure A-1 illustrates the six degrees of freedom about the knee.

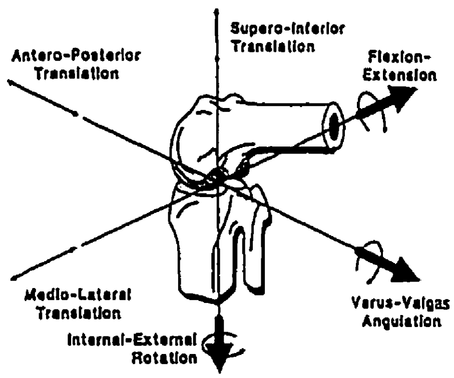

Figure A- 1: Diagram of the six degrees of freedom of movement of the human knee joint (Three rotational and three translational) (Maran's *et al.* 1990)

These terms provide a clinical interpretation of the motion. Translations refer to displacements with respect to the femur. Compression-distraction refers to the entire tibiofemoral joint being shortened or stretched along the longitudinal axis of the bone. It is not the contact areas between the articulating surfaces of the tibia and femur (Lafortune *et al.*, 1992).

Ab/adductory motion is limited to approximately 5° due to the restrictions imposed by the knee's geometry and the collateral ligaments (Frank and Shrive, 1994). Internal-external rotation and flexion-extension are much greater at approximately 35° and 150° respectively. Knee flexion is a combination of the femoral condyles rolling over the tibial plateau and the posterior gliding of the condyles along the plateau. As flexion increases, the translational motion assumes an increasing proportion of knee motion because of the shape of the femoral condyles (Marans *et al.*, 1989).

Functional Bracing

Unlike prophylactic or rehabilitative braces, functional braces provide stability to an unstable knee (Vailas and Pink, 1993). Common to all functional brace construction are the uprights, the hinge and the shell or strapping. Otherwise fabrication and design become the distinguishing marketable characteristics of the brace (Vailas and Pink, 1993). To closely match the kinematics of the normal knee, correct brace design and precise fitting are critical in maintaining the hinges' axis of rotation. Although placement of the axis of rotation is difficult, brace slippage is the primary complaint of wearers. Misalignment could create alterations in forces and moments leading to discomfort from the shearing of the soft tissues underneath the brace or possible abnormal ligament tension (Vailas and Pink, 1993). Key to its rigidity is the straps or shell. The tighter and more rigidly the brace is applied, the better the match for knee motion (Vailas and Pink, 1993).

Early studies reported fewer symptoms of instability with improved athletic performance during brace use although evaluations were anecdotal and subjective in nature (Vailas *et al.*, 1990). Investigations examining the effects of bracing on clinical laxity during static loading reported a reduction but not an elimination of anterior and rotary laxity (Vailas *et al.*, 1990). Current dynamic research purports no performance benefits of bracing. Knees continue to give way during activity and braces fail in situations where high loads are encountered or when the load is applied in an unpredictable manner (Branch *et al.*, 1989; Cook *et al.*, 1989; Vailas *et al.*, 1990; Cawley *et al.*, 1991; DeVita *et al.*, 1992; Beynnon *et al.*, 1992). The lack of supportive evidence for bracing has led investigators to believe that perceived improvement in performances result from a proprioceptive feedback rather than the stabilising effect of a brace (Vailas *et al.*, 1990). Although functional braces may reduce episodes of gross subluxation, they do not prevent abnormal displacements during strenuous activity (Branch *et al.*, 1989). Previous research has indicated that during controlled low load static manoeuvres, bracing appears to be effective in reducing ATT and rotational forces. These forces are thought to accelerate the degenerative joint disease seen in anterior cruciate deficient knees (Branch *et al.*, 1989; Vailas and Pink, 1993). However, braces seem to fail in situations where high loads are encountered or when the load is applied in an unpredictable manner (Branch *et al.*, 1989).

Early Biomechanical Investigations

Six degree of freedom electrogoniometers have been used in order to describe motion of the tibia with respect to a fixed femur during level walking. Marans *et al.*, (1989) reported significant differences in the shape and magnitude of the anterior/posterior translation curve for a group of normal and ACLD subjects. Two distinct patterns were observed: ACLD subjects exhibited increased amplitudes during swing and consistent decreases or absent tibial shifts (Marans *et al.*, 1989).

Of the first to employ high speed cinematography to examine knee kinematics, Tibone *et al.* (1986) failed to note any significant differences between normals and ACLD subjects during walking, running, and stair climbing (Tibone *et al.*, 1986). Branch and Hunter (1990) found that ACLD subjects during the performance of a lateral side step and straight ahead running compensate the way they perform certain athletic manoeuvres. During small athletic manoeuvres such as walking and running, functional braces appeared to be effective and the compensatory mechanisms employed among the ACL deficient group was disrupted (Branch *et al.*, 1989). However, when performing dynamic cutting manoeuvres which stressed the braces to a greater degree, the same support was not evident (Branch *et al.*, 1989).

When higher loads are experienced during athletic activity, the biomechanical effect of stabilising anterior translation with functional bracing is unknown. Improved understanding of joint stability and ACL function during dynamic activities requires measurement during the performance of a dynamic activity. Although previous research has implied variables such as range of joint motion, weight bearing and speed of activity affect ACL, no dynamic research exists to support such assumptions.

Recent Developments and Methodologies

No single methodology exists that produces optimal measuring techniques to record tibiofemoral and patellofemoral joint motion. Although, an abundant amount of data has been collected employing different experimental designs, comparisons between investigations are nearly impossible (McClay, 1990).

New technologies have advanced the means by which tibiofemoral and patellofemoral kinematics have been measured. These include light photography or optoeletric systems (Levens *et al.*, 1948; Lafortune, 1984; McClay, 1990; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994; Reinschmidt, 1996; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b), exoskeletal linkage systems (Marans *et al.*, 1989; Ishii *et al.*, 1997), roentgen-stereo-analysis (RSA) (Lundberg, 1989), and videofloroscopy (Tashman *et al.*, 1995). Although it is beyond the scope of this paper to review all, a brief summary outlining current or promising methods follows. Emphasis is to report on the use of intracortical pins to document three-dimensional tibiofemoral kinematics during the stance phase of walking and running. However, it is important to understand that differences between *in-vivo* investigations particularly in the reporting of frontal and transverse motion may be attributed to variations in the definition of the anatomical co-ordinate system (McClay, 1990). The variations in defining the anatomical co-ordinate system and how tibiofemoral motion is described is detailed later.

External fixator devices

Skeletal motion has been recorded with the use of markers affixed to external fixating devices for patients who have sustained fractures (Angeloni *et al.*, 1993; Andriacchi and Toney, 1995; Cappozzo *et al.*, 1996). This approach has been rarely used as fixators are typically attached to only one segment or patients may not exhibit normal gait due to the injury (Reinschmidt, 1996).

Video Fluoroscopy

Biplanar video fluoroscopy is a promising method for direct measurement of three-dimensional skeletal motion during gait (Tashman *et al.*, 1995). By implanting tantalum pellets, the exact location of bony landmarks for every time frame in the x-ray view is known. This method may help in patient assessments such as testing and knee instability (particularly ACL rupture) during actual movements. However, due to the invasiveness of this technique and exposure to radiation, its use is limited (Reinschmidt, 1996).

Ciné and Video Methods for Recording Tibiofemoral and Patellofemoral Kinematics

No single methodology exists that produces optimal measuring techniques to record tibiofemoral and patellofemoral joint motion although new technologies have advanced the means by which knee joint kinematics can be measured.

Conventional high speed film cameras (typically 100-200 frames per second) and passive markers have been extensively utilised to identify points of interest on the body and quantify human movement (Lafortune, 1984; McClay, 1990; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994; Reinschmidt, 1996; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b). Average measurement errors of 0.03% or 0.6 mm have been reported when recording in a $1 m^3$ volume using film (Lafortune, 1984). However, cinematography has its disadvantages as the process of manual digitisation further increases the chance of errors for determining the joint or marker centre (Winter, 1990; Harris and Wertsch, 1994). For this reason, automated motion analysis systems have replaced film, the most common being passive systems (O'Malley and de Paor, 1993; Harris and Wertsch, 1994). Reflective markers are tracked by an automated multi-camera system and the marker centres are digitised automatically. Although passive systems generally record between 50-60 Hz, 200 Hz cameras can enhance temporal resolution. Errors of 5-6 mm have been reported when recording in a $2m^3$ (Kennedy *et al.*, 1989) volume while other studies found errors of 1-3 mm using similar volumes (Kadaba *et al.*, 1990; Klien and DeHaven, 1995).

The MacReflex motion analysis system, unlike the standard video camera, is an infrared-tracking device designed to detect only reflective markers with high resolution. To assess the accuracy the MacReflex system, a four-segment uniaxial model was specially constructed (Lundberg *et al.*, 1992). Tantalum markers were affixed to each segment of the model and semi-spherical reflective markers were mounted over top. MacReflex recordings and stereo-photogrammetric X-rays (RSA) were taken after a series of perturbations of the different joints. Three-dimensional co-ordinates were calculated for each recording and the data later used for rigid body kinematic analysis. Comparing MacReflex data and the RSA values, Lundberg *et al.*,

(1992) reported standard deviations of less than 0.6° for rotations and less than 0.4 mm for translations when recorded in a volume of 0.25 m^3. When recording in a volume of 0.5 m^3, the corresponding standard deviation values were 1.2° for rotations and 0.9 mm for translations.

Markers and Artefacts

Knowledge about skeletal tibiofemoral kinematics (particularly abduction-adduction and internal/external rotation patterns) are limited since measurements have usually been accomplished by attaching reflective markers to the surrounding soft tissue of the calf and thigh (Ishii *et al.*, 1997). Based on rigid body mechanics, three-dimensional analysis assumes that markers placed on the body represents the position of anatomical landmarks for the segment in question (Nigg and Cole, 1994). However, surface markers may not represent the true anatomical locations resulting in *relative* and *absolute errors* (Nigg and Cole, 1994). Relative errors are movements between markers with respect to each other and are caused by skin movement relative to bone (Ishii *et al.*, 1997). An absolute error is movement of a marker with respect to a specific body landmark (Nigg and Cole, 1994). The local co-ordinate system may not reflect the true geometric relationship of the segment and consequently, considerable questions remain regarding what constitutes normal motion of the knee (Nigg and Cole, 1994; Ishii *et al.*, 1997).

A way to avoid the problem of surface markers is to use invasive markers to directly measure skeletal motion. This provides the most accurate means for determining bone movements (Cappozzo, 1991; Nigg and Cole, 1994). Differences of up to 50% for similar knee angles when comparing tibiofemoral joint kinematics using external and bone fixed markers have been reported (Nigg and Cole, 1994). It appears that skin movement artefacts present the most critical source of error (Cappozzo, 1991).

Intracortical Pin Technique

A pioneer in the use of intracortical pins to study human motion *in-vivo*, Levens *et al.*, (1948) examined the walking patterns of twenty-six subjects in the transverse, sagittal and frontal planes. Threaded stainless steel pins (2.5 mm diameter) were implanted into the cortices of the iliac crest, the tibial tubercle, and the adductor tubercle to negate interference with the Iliotibial (IT) Band. Because of the pins bending, loosening or vibrating during testing, only twelve subjects provided satisfactory data (Levens *et al.*, 1948).

Lafortune (1984) conducted a similar bone pin investigation to examine three-dimensional tibiofemoral and patellofemoral kinematics during normal walking and with shoes modified with varus/valgus soles. Subjects were implanted with Steinman pins (2.5 mm diameter) affixed with target clusters into the adductor tubercle, into the lateral tibial condyle and into the midpatella (Lafortune, 1984; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994). Each triad like the one in Figure A-2 contained four noncollinear spheres, one in the centre and three attached to orthogonal projecting rods (Lafortune *et al.*, 1992).

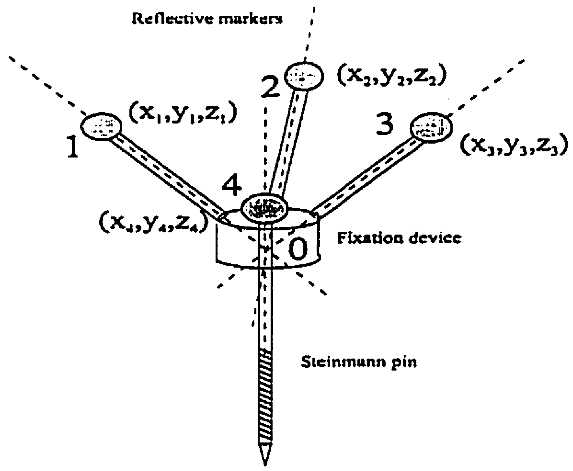

Figure A- 2: Schematic of target cluster used by Lafortune (1984)

To prevent interference with the contralateral leg during walking, the femoral target marker was modified to project anteriorly. Radiographs were subsequently taken with the implanted pins in order to record the position of the markers and define the tibial and femoral anatomical reference points (Lafortune *et al.*, 1992). Since these anatomical landmarks can be identified with great precision, an accurate description of skeletal movement is possible (Reinschmidt, 1996).

Walking trials were recorded using four high-speed cameras and the co-ordinates of each target marker were reconstructed employing a standard linear transform (DLT) (Abdel-Aziz and Karara, 1971). A series of transformation matrices (Lenox and Cuzzi, 1978) resolved the femoral anatomical co-ordinate system into the tibial anatomical co-ordinate system. Subsequent tibiofemoral kinematics was expressed in terms of Cardan or Euler angles with respect to the anatomical co-ordinate system. Rotations and translations were described according to the conventions of the joint co-ordinate system (Grood and Suntay, 1983). Since the location of the anatomical frames of reference were not set to have their origins correspondent, all linear displacements were described relative to the positions of the tibia and femur upon heel strike (Lafortune, 1984; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994).

Using a similar intracortical protocol, McClay (1990) examined whether tibiofemoral and patellofemoral kinematics was altered during running. McClay (1990) analysed two non-injured runners and two subjects who experienced chronic patellofemoral pain. Unlike Levens (1948) and Lafortune *et al's.*, (1992; 1994) studies, the femoral pin was inserted laterally with the knee flexed 45° (McClay, 1990). This reduced the threat of impingement by placing the Iliotibial (IT) band posterior to femoral pin. Additionally, a small longitudinal incision was made through the tissue to minimise restriction. Radiographs with target markers were taken to define the tibial and femoral anatomical co-ordinate systems (McClay, 1990). McClay (1990) employed the exact DLT and transformation algorithms as Lafortune (1992) to describe joint rotations and translations. Knee motion was expressed using the conventions of Grood and Suntay (Grood and Suntay, 1983).

An alternate method was employed using an instantaneous helical axis to describe tibiofemoral joint motion during voluntary swing, normal gait and a pivot manoeuvre (Murphy, 1990). However, the concept of instantaneous helical axes is not widely used in gait analysis since rotations about and translations along a unique spatial axis have no anatomical reference (Reinschmidt, 1996).

Recent investigations compared skin marker and skeletal marker motion during the stance phase of walking and running (Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b). Hoffman bone pins (2.5 mm in diameter) affixed with target markers similar to Lafortune *et al.*, (1992) and McClay (1990) were implanted into the lateral femoral condyle and lateral tibial condyle. A 10-15 mm longitudinal incision into the IT Band reduced impingement with the femoral pin. Six additional surface markers were attached to the thigh and lower leg. Skin and skeletal marker co-ordinates were recorded for one standing trial in a fully extended neutral position and normalised with respect to stance in order to define the tibial and femoral anatomical co-ordinate system. It was assumed that the segmental co-ordinate systems were aligned with the global co-ordinate system during standing. For comparisons with the standing based co-ordinate data, additional roentgen-stereo-photogrammetric x-rays (RSA) were taken to define anatomical references with respect to the tibia and femur (Reinschmidt, 1996; Reinschmidt *et al.*, 1997b). In contrast to neutral standing, RSA enables anatomical meaningful origins to be defined, which allow for joint translation determinations. To ensure accurate representations of skeletal motion, the orientation of the target markers remained fixed throughout the experiment. Employing transformation matrices (Söderkvist and Wedin, 1993), the anatomical femoral co-ordinate system was resolved into the anatomical tibial co-ordinate system, similar to the calculated standing trials. Knee motion based on external (thigh, shank) and skeletal (femur, tibia) markers were expressed in terms of Cardan angles using the conventions of Grood and Suntay (Grood and Suntay, 1983).

Reinschmidt (1996) presented rotational data based on neutral standing that focused on differences between external and skeletal based kinematics. Since the femoral pin appeared stable for the remaining subject, rotations and translations derived from RSA and neutral standing were presented for this individual only.

Roentgen-stereo-photogrammetric analysis

Roentgen-stereo analysis (RSA) is used to calculate three-dimensional positions of bony landmarks identified in two or more radiographic pictures. Since anatomical landmarks can be identified with great precision, an accurate description of skeletal movement is possible when employing either Euler angles or the Screw axis with respect to a body oriented co-ordinate system (Reinschmidt, 1996).

To define both tibial and femoral anatomical reference frames, Lafortune *et al.*, (1992) McClay (1990) and Reinschmidt (1996; Reinschmidt *et al.*, 1997a) all used identical RSA based definitions. Briefly, stereo-radiographs were taken from the lateral and anterior views. Both femoral and tibial target markers were digitised in addition to anatomical points of interest (figure 3.2). The deepest point of the intercondylar groove was chosen as the origin for the femoral co-ordinate system. The longitudinal axis passed through the origin and was directed superiorly and parallel to the long axis of the femur. The medio-lateral axis progressed along a line connecting the most distal points on the medial and lateral femoral condyles, passed through the origin and perpendicular to the longitudinal axis. The remaining axis was calculated using the cross product of the two defined unit vectors. The origin for the tibial co-ordinate system was located on the most proximal point of the medial intercondylar eminence. A line parallel to the longitudinal axis of the tibia and passing through the origin was used to define the proximal-distal axis. The medio-lateral axis progressed along a line through the estimated centres of the medial and lateral tibial articular surfaces, passing through the origin and perpendicular to the longitudinal axis. The remaining anterior-posterior axis was calculated using the cross product (Lafortune, 1984; McClay, 1990; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994; Reinschmidt, 1996; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b).

Analytical methods to Quantify Joint Motion

To date, different methods have been used to describe and quantify the three-dimensional kinematics of the lower limb *in-vivo* during various movements. They include helical axes (Woltring, 1994), finite helical axes descriptors (Lundberg, 1989), instantaneous helical axes (Murphy, 1990), and joint co-ordinate systems based on local anatomic landmarks (Grood and Suntay, 1983; Chao *et al.*, 1983). The helical axes method is the most unique, in that general rigid body motion is described as a screw movement. Clinical interpretation is often difficult since instantaneous helical axes are not referenced to an anatomical segment. Conversely, the joint co-ordinate system is the most common method and uses Cardanic (or Euler) angles with respective translations to describe joint motion about axes defined in the anatomical segments (Reinschmidt, 1996). Each method accurately describes the relative skeletal motion in 6 degrees of freedom, their differences being how the motion is partitioned (Reinschmidt, 1996). Although these methods have advantages and disadvantages, each is dependent on the research question.

To define the anatomical co-ordinate system, methods include neutral standing (Areblad *et al.*, 1990; Nigg *et al.*, 1993; Reinschmidt, 1996; Moseley *et al.*, 1996; Reinschmidt *et al.*, 1997b), roentgen-stereo-photogrammetric analysis (Lafortune, 1984; McClay, 1990; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994; Reinschmidt, 1996; Reinschmidt *et al.*, 1997a), and relationships between bone embedded reference frames and external markers placed on anatomical landmarks (Cappozzo *et al.*, 1995). After having established the anatomical reference frames, a co-ordinate transformation matrix consisting of three rotational and three translational degrees of freedom is employed to resolve the femoral anatomical co-ordinate system into the tibial anatomical co-ordinate system (Lenox and Cuzzi, 1978; Spoor and Veldpaus, 1980; Söderkvist and Wedin, 1993).

For gait analysis in clinical settings, the most commonly used co-ordinate system is the "joint co-ordinate" system (Grood and Suntay, 1983). This system calculates 3D joint attitude parameters as well as joint translations by partitioning general joint motion into 6 familiar anatomic motions based on Cardan or Euler angles (Reinschmidt, 1996). One joint axis is fixed to the proximal segment, the other joint axis fixed to the distal segment, and the remaining floating axis normal to the two fixed body axes. According to the conventions described by Grood and Suntay (1983), flexion/extension and medial-lateral shift occurred around the fixed medio/lateral femoral axis, ab/adduction and anterior-posterior drawer around the floating axis and internal/external knee rotation around fixed tibia proximal/distal axis.

Tibiofemoral Motion

Levens *et al.*, (1948) identified general patterns of internal and external rotations for the pelvis, femur, and tibia from heelstrike to midstance and from midstance to toe-off respectively. Greater internal and external rotations were evident for the distal segments than proximal ones. More significantly, Levens *et al.*, (1948) first objectively documented the "screw home mechanism". They found that as the knee locked into extension, the femur internally rotated with respect to the tibia. Conversely, as the knee unlocked as during flexion, the femur externally rotated. General tibial motion was described as a relative inward rotation of approximately 3.5° with respect to the femur between late swing (locked knee position) and midstance. From midstance until toe-off, no relative rotations were observed although a slight outward rotation of 1.5° beyond full weight bearing was noted followed by a slight internal rotation of approximately 0.5°. A further outward rotation of about 3.5 ° was observed as the foot approached toe-off (Levens *et al.*, 1948).

Walking investigations

Flexion/extension:

Flexion/extension patterns are the largest component of total knee motion during walking. Lafortune *et al.*, (1992) and Reinschmidt *et al.*, (1997a) skeletal based flexion/extension curves compared favourably both in shape and magnitude. At heel strike, the knee is slightly flexed between 0-10° and continually increased (15°-20°) until approximately 15-20% of stance. Then extension occurred just short of full extension (defined to be 0°) at about 60% of support. Knee flexion follows again through toe-off. The total range of motion of the tibiofemoral joint during stance is approximately 40° degrees (Lafortune *et al.*, 1992; Reinschmidt *et al.*, 1997b). Reinschmidt *et al.*, (1997b) reported little differences between skin and skeletal based kinematics as the shape of flexion/extension patterns were in general agreement across subjects.

Ab/adduction:

Lafortune *et al.*, (1992) found little or no ab/adductor movements during stance. From heelstrike to shortly before toe-off, the tibiofemoral joint remained abducted approximately 1.2°. Four subjects demonstrated abduction throughout whereas one subject had a constant slightly adducted position. Conversely, Reinschmidt *et al.*, (1997b) found no general patterns across subjects, however, greater abductory-adductory ranges of motion were found; varying between 5°- 10°. The dissimilarity between investigators may be attributed to differences in defining the anatomical co-ordinate systems. Lafortune *et al.*, (1992) employed roentgen-stereo-photogrammetric analysis whereas Reinschmidt *et al.*, (1997b) utilised a neutral standing trial. Unphysiologically high ab/adductor patterns found in one subject was attributed to cross talk with knee flexion/extension, a characteristic of alignment problems of the anatomical co-ordinate system. The authors further reported poor agreement in the shape of skin and skeletal based ab/adductor curves across subjects.

Internal/external rotation:

Lafortune *et al.*, (1992) reported two internal rotations averaging slightly less than 5° across all subjects; one occurring from heelstrike to 25% of stance and one occurring during the last 30% of stance. During mid stance, the tibiofemoral joint remained close to neutral position (0°). Using modified footwear that forced the foot into extreme pronation and supination, Lafortune *et al.*, (1994) found increased initial tibial rotations for the valgus-wedge shoes than with the varus wedge shoes immediately following heelstrike. Overall, no discernible differences in the patterns of tibiofemoral internal/external rotation were evident when wearing modified shoes. Angular patterns and translations were altered by less than 1 and by 2 mm respectively. This suggests increased internal or external tibial rotations may be resolved at the hip joint in healthy individuals with changes at the tibiofemoral joint barely detectable (Lafortune *et al.*, 1994).

Reinschmidt *et al.*, (1997b) found greater intersubject variability than the findings of Lafortune *et al.*, (1992). Subjects either demonstrated initial internal rotations or external rotations with the skeletal markers. Their overall ranges of motion varied from 5° to over 10°.

Poor agreement was found across subjects in the shape of skin and skeletal based internal/external rotations. Internal/external rotations derived from skin markers with respect to the global co-ordinate system were also calculated (Reinschmidt, 1996). Consistent internal tibial rotations were observed across subjects from touchdown to about 25% of stance; which were not present at all or to a much lesser extent for skeletal motion (as stated above). Initial femoral rotations were also present and it appears to be matched with initial tibial rotations. Table B-1 shows detailed angular data for the tibiofemoral joint during walking for comparisons across investigations.

Skin marker based kinematics, particularly rotations in the frontal and transverse plane, must be interpreted with caution. Although surface marker flexion/extension patterns were in general agreement with pin derived data, the poor agreement between skin and skeletal ab/adduction and internal/external rotation curves suggests skin marker based kinematics may not reflect true tibiofemoral motion (Reinschmidt, 1996). Reinschmidt found the relative movements between skin markers and the underlying bone can be as high as the motion itself. Therefore, interpretations of the results may lead to incorrect conclusions.

Linear Kinematics

Since the location of the anatomical frames of reference were not set to have their origins correspondent, Lafortune *et al.,* (1992) described all linear displacements relative to the positions of the tibia and femur at heelstrike. Lafortune *et al.,* (1992) discovered a distinct relationship between knee flexion-extension and tibial translations along all three femoral orthogonal axes. About the mediolateral axis, tibial shifts closely matched the patterns of knee flexion/extension. An initial 2.3 mm medial tibial shift occurred when the knee flexed early during stance, followed by a 1.5 mm lateral shift as the knee extended during the middle part of stance (Lafortune, 1984; Lafortune *et al.,* 1992). Regarding anterior/posterior drawer (movements along the floating axis), the tibia was drawn posteriorly when the knee flexed and it moved anteriorly during extension. Posterior drawer amounted to 3.6 mm during the first half of stance while extension was associated with a maximum anterior displacement of 1.3 mm past the neutral position, defined as 0 mm (Lafortune, 1984). After heelstrike, a maximum distraction of 3.2 mm during flexion occurred followed by a 0.2 mm compression accompanying knee extension. Table B-2 shows detailed linear data for the tibiofemoral joint during walking.

Running Investigations

It should be noted that the Lafortune *et al's.*, (1992) data is specific to walking and the characteristics may be different when compared to McClay's (1990) and Reinschmidt *et al's.*, 1997a) investigations. Furthermore, the calibration procedures utilised by McClay (1990) and Reinschmidt *et al.*, (1997a;1997b) were not optimal, reducing the accuracy of spatial reconstructions. McClay calibrated without the use of a calibration frame and Reinschmidt employed a less than adequate calibration frame due to its size and small number of control points. The discrepancies between McClay's unpublished data and Reinschmidt's investigations may result from differences in defining the tibial and femoral anatomical co-ordinate systems. The anatomical co-ordinate system employed by McClay was based on a roentgen-stereo-photogrammetric analysis whereas Reinschmidt utilised both a neutral standing trial and RSA method. Finally, differences in running styles may also account for some variations.

The reader should also note that Reinschmidt *et al.*, (1997a; 1997b) presented rotational data for three subjects based on neutral standing focusing on differences between external and skeletal based kinematics and skeletal marker intrasubject variability. Comparisons between RSA and neutral standing rotations and translations are based on data from one subject since the femoral pin appeared stable for only this subject.

Flexion/extension:

During running, McClay (1990) found the knee generally flexed 10° - 20° at heelstrike to around 30° - 40° approximately 40% in stance. It then extended shortly before toe-off and flexed in preparation for swing. McClay reported flexion/extension ranges of motion of 21° and 26° for the normal group and the patellofemoral pain group (PFP) respectively. Normals had greater flexion angles at heelstrike and remained in greater flexion throughout compared with the PFP subjects. Peak flexion and extension velocities were 20 - 25% greater for the pathological group. McClay suggested this may be a contributing factor in movement pathology due to the greater amounts of strain experienced by the soft tissue (McClay, 1990)

Comparing skin and skeletal kinematics derived from neutral standing trials, Reinschmidt found flexion/extension patterns similar in shape and magnitude across subjects with a systematic offset between the two curves (Reinschmidt, 1996; Reinschmidt *et al.*, 1997a). The authors suggested changes in muscle activation between neutral standing and running caused the shift between skin markers and underlying bone (Reinschmidt, 1996; Reinschmidt *et al.,* 1997a). The skeletal flexion/extension curves were similar in shape and magnitude across subjects and corresponded well with McClay's (1990) investigation. However, small differences in knee position at heelstrike were evident across subjects ranging from 0° to 15°. Reinschmidt attributed these differences to alignment uncertainties in defining the anatomical co-ordinate systems (Reinschmidt, 1996). The RSA and neutral standing skeletal flexion/extension curves agreed well in shape and amplitude however a consistent shift between the curves was evident.

Ab/adduction:

In contrast to Lafortune *et al.*, (1992), McClay (1990) found clear ab/adduction patterns across subjects during stance phase. From heelstrike to about 40% stance, the normals' tibia adducted 6° followed by a gradual abduction of 8° until the end of stance. Comparing ab/adduction curves between subjects and groups, the patterns agreed well although peak adductory amplitudes were higher for the injured group. The total range of motion was similar averaging approximately 8° across both groups.

Reinschmidt's *skeletal* ab/adduction patterns and magnitudes were in total disagreement with McClay's data (Reinschmidt, 1996; Reinschmidt *et al.,* 1997b). Ab/adduction patterns varied considerably among subjects. The contradiction between Reinschmidt and McClay may be attributed to their definition of the anatomical co-ordinate systems based on neutral standing and RSA respectively.

Reinschmidt also found poor agreement between *skin* and *skeletal* ab/adduction patterns since the external markers did not reflect "true" skeletal movement patterns (Reinschmidt, 1996; Reinschmidt *et al.,* 1997a). The differences did not appear to be systematic which suggested ab/adductory kinematics were subject dependent.

Reinschmidt's *skeletal* ab/adduction curves derived using both the neutral standing trial and RSA agreed well although a constant shift between the two average curves was evident (Reinschmidt, 1996; Reinschmidt *et al.,* 1997a). Interestingly, Reinschmidt's single subject RSA curve totally disagreed with McClay's (1990) RSA data although both employed the exact femoral and tibial anatomical co-ordinate system (Reinschmidt, 1996; Reinschmidt *et al.,* 1997a).

As stated earlier, ab/adduction ranges of motion is limited to approximately 5° (Frank and

Shrive, 1994). Since both McClay (1990) and Reinschmidt (1996) reported motion sometimes exceeded 5°, Reinschmidt suggested that "true" ab/adduction patterns may have not been reflected with these techniques. Since the patterns of ab/adduction are similar to flexion/extension, Reinschmidt (1996) speculated that the small ab/adductory magnitudes coupled with problems in aligning the anatomical co-ordinate system resulted in cross talk that masked the actual motion.

Internal/external knee rotation:

McClay (1990) found similar internal/external patterns across subjects and between conditions although the total range of motion varied slightly between normals and the Patellofemoral Pain (PFP) group respectively. From touchdown to midstance, the knee internally rotated which was followed by an external rotation although slight differences in magnitudes were evident between groups. Furthermore, the PFP group demonstrated a delay in reaching peak internal rotation of 25ms.

Reinschmidt (1996) found poor agreement between skin and skeletal internal/external patterns, an indication that motion was highly subject dependent. Based on neutral standing trials, skeletal internal/external patterns were fairly similar across subjects although the magnitudes varied between subjects. From heelstrike to midstance, subjects demonstrated either a pronounced initial internal rotation varying between 7°-9° or a small rotation of 2°. During the later half of stance, the knee externally rotated. The patterns and magnitudes compared favourably with McClay's (1990) investigation although in that study the magnitudes of internal rotation were higher than external rotation during the second half of stance. Skeletal internal/external curves based on both the neutral standing trial and RSA agreed well but an obvious offset between the two average curves was evident. Furthermore, Reinschmidt found noticeable differences for internal knee rotations between the two co-ordinate systems not evident for the other rotations. Initial internal knee rotations based on RSA and neutral standing trials averaged 11° and 7° respectively. Small deviations and inconsistencies in defining the anatomical co-ordinate systems may account for the differences across subjects. Table B-3 summarises detailed angular data across studies for the tibiofemoral joint during running.

Ab/adduction and internal/external rotation curves were similar in shape to flexion/extension patterns although amplitudes were lower. It was speculated that the small ab/adductory and internal/external magnitudes coupled with problems in aligning the anatomical co-ordinate system resulted in cross talk and masked the actual motion. Conversely, others studies reported that knee rotations may be coupled; for instance, as the knee undergoes flexion an internal rotation takes place (Blankevoort *et al.*, 1988; Lafortune *et al.*, 1992).

From the lack of studies delineating these patterns, it is difficult to estimate how much of the non-primary rotations (ab/adduction, internal/external rotations) are "real" or how much can be attributed to alignment problems of the anatomical reference frames (Reinschmidt, 1996). It has also been suggested that tibiofemoral joint kinematics derived from Cardan angles and described according to the conventions of the joint co-ordinate system (Grood and Suntay, 1983) may not be appropriate to determine knee rotations other than flexion/extension. More research is required to establish a reliable co-ordinate system to enable valid comparisons across subjects.

Linear Kinematics

Similar to Lafortune *et al.*, (1992), McClay (1990) associated flexion with tibial distraction and translations both medially and posteriorly with the opposite being true for extension. Although the magnitudes and patterns between investigations vary, the differences can be attributed to differences in locomotor activity or differences in the placement of the anatomical axes. Since the origin of the femoral and tibial anatomical co-ordinate systems are some distance apart from each other, Reinschmidt reported translations as changes in movement (Reinschmidt, 1996; Reinschmidt *et al.*, 1997b). Conversely, Lafortune *et al.*, (1992) and McClay (1990) described linear displacements as absolute values relative to the positions of the tibia and femur at heel strike.

Anterior/Posterior Drawer

The shapes of the McClay's (1990) anterior/posterior curves were fairly similar across conditions and they corresponded well with respect to Lafortune *et al.*, (1992). Higher magnitudes were evident during running. At heelstrike, McClay reported the origin of the tibial anatomical reference frame was posteriorly placed with respect to the femoral origin. A further posterior displacement occurred during the first 25-35% of stance when the knee flexed after which the tibia moved anteriorly. Throughout stance, the tibial reference point was more anterior for the pathological group in comparison to the normals. They also exhibited greater anterior drawer but McClay (1990) was unable to account for these differences between conditions.

During the first 5% of stance, Reinschmidt (1996) observed negligible translations which was followed by a posterior tibial displacement of 4 mm between the origins of the tibial and femoral reference frames. From about 40% to 80% of stance, the tibia moved anteriorly (5mm) followed by a fast posterior displacement towards the end of stance. RSA based anterior/posterior drawer patterns were similar to the patterns reported by McClay (1990).

Medial Lateral

All subjects in McClay's (1990) study exhibited a medial translation during the first half of stance that was followed by a lateral translation until toe-off. This pattern is consistent with the findings of Lafortune *et al.*, (1992).

The least amount of translationary movement according to Reinschmidt (1996) was medio-lateral shift during stance. No consistent patterns were evident during the first 15% of stance after which the tibia underwent a lateral shift with respect to the origins of the tibial and femoral reference frames. From 40% to 80% stance, the tibia shifted medially followed by a lateral shift during the last 20% of stance. Reinschmidt's (1996) RSA based medio-lateral patterns were generally in the opposite direction as McClay (1990) reported. This discrepancy may be related to the discrepancies found for the ab/adduction motion.

Distraction/compression

Although the articular surfaces come together during loading as the knee flex, the selected origins distract as a result of the curvature of the femoral condyles during flexion (McClay, 1990). Distraction continued until midsupport followed by compression.
Reinschmidt (1996) observed no compression/distraction during the initial 10% of stance. From 10% to 40% of stance, a 5.6 mm distraction was first noted which was followed by an even larger compression of 6.8 mm between the origins of the two co-ordinate systems. During the final 20% of stance, a distraction movement of 2.8 mm was observed. Reinschmidt's (1996) RSA based distraction/compression drawer patterns derived for the single subject were similar to the patterns reported by McClay (1990). A summary of the linear data compiled across studies for the tibiofemoral joint during running is found in Table B-4.

Similar to Lafortune *et al.*, (1992) and McClay (1990), tibiofemoral translations exhibited a striking similarity with knee flexion and extension behaviour. Reinschmidt (1996) claims all translations are dependent on the rotations since tibial and femoral reference frame origins do not reflect an "average" knee joint centre. Cross talk would register a translation even though a pure rotation would take place. Blankevoort *et al.*, (1988) suggested meaningful distances be calculated between points embedded in the two bodies (e.g. ligament insertion sites) which would provide more comprehensive and physiological meaningful translations than translations calculated along the axis of a joint co-ordinate system. From the RSA radiographs, the location of the femoral and tibial insertion sites of the ACL were digitised in an attempt to calculate the distance between these points during the stance phase of running (Reinschmidt, 1996). Although translations were smaller, the measurements were within the range of measurement error therefore the data was not presented.

Screw Home Mechanism

The screw home mechanism is generally defined as a combination of knee extension and external rotation of the tibia about the femur. Tibiofemoral joint motion can best be described as spiral or helicoid during flexion and extension (Nordin and Frankel, 1989). This spiral motion occurs because the medial femoral condyle is longer than the lateral. As the tibia glides on the femur from the full flexion to full extension, it descends and then ascends the curves of the medial femoral condyle and simultaneously rotates externally varying between 0° - 14° (Lafortune, 1984; Nordin and Frankel, 1989). The motion is reversed as the tibia moves back into the fully flexed position. Such a mechanism provides more stability to the knee than would a simple hinge configuration.

Lafortune *et al.*, (1992) reported that the knee approached maximal extension twice; once during stance and once during swing. Twice during stance, the tibiofemoral joint rotated internally averaging slightly less than 5° across all subjects; one from heelstrike to 25% of stance and one during the last 30% of stance. When the knee was loaded from mid-support until just before toe-off, the tibiofemoral joint remained close to the neutral position (0°) and exhibited no external rotations although extension occurred. During the unloaded phase of the gait cycle, Lafortune *et al.*, (1992) reported external tibiofemoral rotations (9.4°) for most of the cycle when the knee either flexed or extended. The results do not support the concept of the screw home mechanism during locomotion.

McClay (1990) found general agreement that flexion was associated with tibial internal rotation and adduction while extension was associated with the external rotations. During the last half of stance as the knee was achieving maximum knee extension near toe-off, external rotation dominated. This suggests support for the screw home mechanism although McClay could not substantiate this as the subjects did not reach full extension at any time during the support phase.

Although Reinschmidt (1996) did not report on the screw home mechanism, analysis of the skeletal marker data by this author suggests some support for this phenomena. Patterns of tibial internal/external rotations from Reinschmidt's (1996) walking investigation show more inter-subject variability than the findings of Lafortune *et al.,* (1992). During flexion, two subjects demonstrated minimal or clear initial internal rotations of approximately 2° and 5° respectively from heelstrike to 25% of stance. One subject initially externally rotated 4° during flexion. Upon extension, patterns of external rotations were evident across subjects. As toe-off neared, internal rotation dominated. Reinschmidt (1996) also calculated tibial and femoral internal/external rotations with respect to the global laboratory co-ordinate system. Graphically, consistent internal tibial rotations were evident across subjects from heelstrike to approximately 25% stance. External rotations are apparent until extension followed by internal rotations prior to toe-off. These results support the generally accepted paradigm of internal rotation at and shortly after touchdown. However, internal tibial rotation appears to be matched by internal femoral rotations. While running, the same two subjects demonstrated clear internal rotations from heelstrike until midstance whereby external rotations were observed to peak extension. Prior to toe-off, internal rotations began. Internal/external rotations calculated from skin markers must be interpreted with caution as the error introduced as a result of skin movement artefact can be as high as the motion measured.

Patellofemoral Joint Motion

Because of measurement difficulties, little is known about patellofemoral joint motion. Lafortune (1984) was among the first to investigate its motion *in-vivo* during locomotion. Although data from five subjects were collected, two were discarded due to fixation difficulties of the patella pins.

In describing the angular displacements of the patellofemoral joint, the joint co-ordinate system was similarly employed. Flexion/extension occurred around the X_F femoral fixed axis, internal/external rotation around the Z_P body fixed axis, and ab/adduction around the floating F. Patellofemoral translations were resolved into X_F, Y_F, and Z_F components (shift, run and glide) respectively along the femoral axis. Absolute position and displacements were be reported because of the relatively small size of the patella and because translations were resolved along the femoral anatomical axes. Lafortune (1984) reported patellofemoral joint motions using the tibiofemoral flexion/extension patterns as a time reference. The mean values from two trials for each of the three remaining subjects was presented.

Angular kinematics

Lafortune (1984) observed patellofemoral flexion/extension patterns and ab/adduction patterns were consistent across subjects and their ranges of motion were smaller than the tibiofemoral joint. The patellofemoral joint exhibited more hyperextension which Lafortune attributed to the shape of the articular surface of the femoral condyles. The femoral condyles bulge anteriorly with respect to the femoral shaft causing the patella to be hyperextended when the tibiofemoral joint is in extension. Average sagittal patellofemoral joint motion exhibited the same general trend as the corresponding tibiofemoral patterns; they simultaneously flexed and extended with coincident peak angular values during stance and swing. At heelstrike, the patella was located 11.8 ° of extension and flexed to a neutral position (0°) until midstance. A positive value indicates extension while negative means flexion. From midsupport to maximal tibiofemoral extension, the patellofemoral joint reached 12.5°. Thereafter, the tibiofemoral joint flexed until toe-off and the patella was placed -16.1°. Average ab/adduction patterns showed the patella to be neutrally aligned (adducted 0.6°) immediately preceding heelstrike after which it began to adduct (2°) and remained adducted until approximately 50% of stance. Then it abducted reaching 6.2° at toe off. It was also found that patellofemoral ab/adductory motion was much larger than for the tibiofemoral joint. The rotational movements were highly variable with one of the subjects exhibiting a different motion from the other two. In general, all patellae remained externally rotated throughout the stance phase. At heelstrike, the patella was in about 5° of external rotation. Two subjects initially internally rotated until approximately 50% of stance then began to externally rotate to 8.7° of extension as the tibiofemoral joint reached maximal extension. Internal rotation followed with the final position at toe-off being approximately 6° of external rotation.

McClay (1990) also reported that patellofemoral flexion/extension patterns closely resembled the tibiofemoral motion. This is likely due to the patella's attachment proximally and distally to the femur and tibia respectively (McClay, 1990). Although the ranges of motion were smaller, sagittal motion comprised the largest component of angular displacements. Unlike Lafortune (1984), patellofemoral flexion/extension patterns were more variable across subjects than those of the tibiofemoral joint. From heelstrike until approximately 40% of support, the patella flexed followed by extension peaking at 90% support. The patella then flexed again through toe-off. The pathological group demonstrated less flexion at heelstrike and less peak flexion at midstance but greater ranges of motion compared to normals. Velocity at heelstrike, peak flexion and peak extension velocities were all higher for the pathological subjects. Similar to Lafortune's (1984) investigation, McClay (1990) found coincident peak tibiofemoral and patellofemoral flexion values.

Most subjects demonstrated fairly similar ab/adduction patterns. In contrast to Lafortune (1984), the patella initially abducted from heelstrike to midstance followed by adduction until approximately peak stance extension. An additional abductory phase occurred prior to toe-off. No clear differences were evident between groups although one PFP subject exhibited an opposite pattern. As McClay reported, when the tibia internally rotated during tibiofemoral flexion, the tibial tubercle moved medially. This functionally decreased the Q angle placing the patella more in adduction. However, the vastus medialus acts medially at the superior pole of the patella that causes it to abduct. It is likely that the frontal plane movements are a balance between these two factors and the variability between subjects is a result of anatomical and neuromuscular differences.

McClay (1990) found similar internal/external rotational patterns across subjects and between conditions although offsets were evident. At heelstrike, two subjects had the patella in external rotation, one in internal rotation and one neutrally aligned. Similar to Lafortune (1984), subjects landed with the patella externally rotated and remained in this position until 50%-75% stance when further external rotation occurred. In general, all subjects exhibited very little rotational movements but two externally rotated from approximately 75% contact until toe-off. No direct relationship between internal and external rotation of the patella with the internal and external rotation of the tibia was evident (McClay, 1990). A summary of the patella angular linear data can be found in Table B-5.

In general the patterns of this joint were more variable between subjects. During tibiofemoral flexion, the patella flexed, adducted, posteriorly and medially translated, and distracted with respect to the femur. The patella remained anterior, superior and medial to the femur throughout stance.

The differences in patellofemoral joint motion between Lafortune (1984) and McClay (1990) can be attributed to the differences in tibiofemoral joint motion during walking and running. McClay reported greater knee flexion angles from heelstrike until peak extension compared to Lafortune's study. At toe-off, the knee was in greater extension during running. With the knee in greater extension during walking, the patella is pulled by the quadriceps and it hyperextends with respect to the femoral shaft. This would explain the greater patellofemoral extensions from heelstrike to toe-off observed in Lafortune's (1984) walking investigation compared to the patella being overall flexed during running in McClay's (1990) study.

Linear kinematics

Lafortune (1984) reported the patterns of patellar shift (medial/lateral) exhibited the greatest variability between subjects with uneven lateral displacements occurring throughout stance. Overall, Lafortune found that the patella was medially, anteriorly, and proximally placed with respect to the femoral anatomical reference frame throughout gait. In general, as the knee flexed, the patella displaced laterally, posteriorly and distally. Conversely, it moved in the opposite direction as the tibiofemoral joint extended. Furthermore, anterior/posterior, medial/lateral, and proximal/distal translations followed closely the patterns of tibiofemoral flexion/extension.

McClay (1990) reported patellar translations along the floating axis were fairly similar in shape and magnitude across subjects although one was offset anteriorly. At heelstrike, all patellae were initially positioned anterior to the femur and remained stationary in this position for approximately 50 ms. During tibiofemoral flexion, the patella displaced posteriorly which was followed by an anterior movement during extension. The pathological group exhibited little or no posterior movement following heelstrike compared to the normal group. These differences may be accounted for since the PFP group was in less tibiofemoral flexion throughout stance than the NL group (McClay, 1990). Lafortune (1984) reported similar posterior movements during flexion although posterior translations began immediately following initial heel contact.

Total range of motion for the normal group was almost twice that of the PFP group respectively.

Most subjects demonstrated initial medial translations of the patella with respect to the femur followed by a lateral translation. However, one exhibited very little mediolateral translation. In general, the patellae were positioned medial to the femur at heelstrike and throughout stance. One PFP subject had the patella slightly lateral to the femur at heelstrike but immediately translated medially upon contact. The most notable translatory finding was the PFP group exhibited medial-lateral excursions approximately 2.75 times than that of the normal group.

Proximal-distal translations were fairly similar in shape and magnitude across subjects and resembled tibiofemoral and patellofemoral flexion/extension patterns. In all cases, the patellar origin remained proximal to the femoral origin throughout support. Generally, the patella distally translated at heelstrike and peaked 40% into stance. It then moved proximally peaking at 90% of support followed by a distal translation through toe-off. These findings are consistent with those of Lafortune (1984) although he reported the patella initially moved proximally following heelstrike. Thereafter, proximal/distal translations closely followed tibiofemoral flexion/extension patterns. A summary of the linear data can be found in Table B-6.

Effect of Bone Pins

In these studies, none of the subjects experienced pain and/or significant discomfort during the experiments; all reported being able to move their knee freely despite pin implantation (McClay, 1990; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b). Subsequent to the surgery, no problems were associated with either the femoral and tibial insertion sites and most engaged in normal activities two weeks following the experiment.

In order to quantitatively assess whether the pins affected knee kinematics, Reinschmidt (1996) compared skin marker kinematics for walking and running with and without bone pins. During walking, differences did not exceed 2.1° for ab/adduction, 4.8° for internal/external rotation and 4.5° for flexion extension whereas differences were < 3° for all rotations for running. Although these differences may be considered substantial, the similarity in the shape and amplitude of the curves suggests the bone pins did not affect walking and running styles. What is immediately evident is the systematic shift between the pin and non-pin curves for both walking and running trials. These shifts were attributed to the different standing trials for the non-pin and pin trials causing slight differences in defining the neutral position.

Sources of Error in Knee Motion Measurements

Accuracy of Spatial reconstruction

To estimate the accuracy of the calculated camera constants, Lafortune used the constants to repredict the spatial location of the control points in both the global and radiograhic reference frames (Lafortune, 1984; Lafortune *et al.,* 1992). Lafortune reported errors of 0.5 mm for the global reference frame and errors less than 0.4 mm for the radiographic reference frame. This represented an average error of less than 0.03 % in the experimental area.

To spatially reconstruct the data, McClay (1990) employed a modified version of the Simultaneous Multiframe Analytic Calibration method (SMAC). The target clusters themselves served to "self-calibrate" the experimental area and allow for reconstruction of their global co-ordinates (Woltring *et al.,* 1989). Output from the SMAC procedure provided position components for each camera with respect to the reference frame (femoral target cluster), co-ordinates of the principal point, principal distances and the parameters which evaluated the degree of orthogonality of the image axes. Additionally, mean and standard deviations of the six degrees of freedom (positional and rotational) of the reconstructed clusters were provided to indicate the accuracy of the procedure. Following SMAC, camera parameters were converted to conventional DLT parameters and reconstructed using a normal DLT process, which resulted in three-dimensional co-ordinates in an arbitrary global system.

To assess the accuracy of the entire procedure (including the SMAC), mean inter-LED distances (across 15 trials) were calculated for each cluster. Mean ILED distance was measured at approximately 70 mm and the error between the calculated and measured distance was 0.97 mm ± 0.86. This represents approximately a 1% error in the reconstruction process. ILED distance varied during stance with a mean range of 0.178 cm ± 0.107 which represents a 2% error. This mean range is independent of the reconstruction process and most likely is the result of the LED's change in position with respect to the camera as the lower limb moved through the experimental field. The consequence of these errors translates to angular and linear uncertainties of 1.28° and 2.62 mm respectively. One must exercise caution when drawing conclusions when displacements are under 2.6 mm or when angular excursions are less than 1.3°.

Reinschmidt reported that discrepancies between skin and skeletal kinematics may be masked by inaccuracies in determining the spatial position of markers moving outside the calibrated volume. The proximal thigh markers (surface) were typically outside the calibrated volume although the corresponding skeletal (femur) markers were within the calibrated volume. To assess the accuracy of the DLT calculations, the means of the spatial reconstruction residuals for each external and skeletal marker was calculated for both walking and running trials. The largest distribution between the two was used to estimate the accuracy in determining the knee motion. Skin marker residuals for the greater trochanter appeared consistently higher and therefore were excluded from the calculations in determining knee motion. Residual errors for the remaining markers were approximately 2.5 mm.

Reinschmidt (1997a) reported the 2.5 mm residuals yielded a 2° error in orientation of the skeletal segments and a 1° error for the external segments. It was concluded that differences between skin and skeletal knee rotations in excess of 2° cannot be attributed to inaccuracies of the motion analysis system rather to the combined effect of the skin marker movement artefacts acting at the shank and thigh.

Segmental Error Analysis

To determine the contribution of the skin movement artefact between skin and skeletal based kinematics Reinschmidt performed a *segmental error analysis* (Reinschmidt, 1996; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b). The rotations of the thigh (skin) with respect to the tibia (bone) was subtracted from skeletal tibiofemoral rotations in order to determine the error caused by skin (thigh) movement artefacts. Similarly, shank (skin) motion relative to femur (bone) motion was determined by subtracting the femur-shank based knee rotations from the skeletal femoral-tibial based motion. Errors due to skin movement artefact at the shank were small (< 3° for ab/adduction and < 2° for flexion/extension) with errors not exceeding 5° for all subjects and rotations during walking and running respectively. Errors at the thigh were consistently higher.

Anatomical Co-ordinate System and Cross Talk

When measuring three-dimensional motion *in vivo*, the choice of anatomical co-ordinate systems is of great importance (Reinschmidt, 1996). Cardan angles and the corresponding translations calculated using the Joint Co-ordinate System are highly susceptible to alignment errors and uncertainties in defining the anatomical co-ordinate system (Reinschmidt, 1996). Ramakrishnan (Ramakrishnan and Kadaba, 1991) manipulated the anatomical thigh co-ordinate system along the longitudinal axis and reported no effects on flexion/extension but significant errors in ab/adduction and internal/external knee rotations (Reinschmidt, 1996). The problem of defining the anatomical co-ordinate system makes comparisons across subject and studies difficult since subtle differences may easily be caused by small deviations in anatomical reference alignment (Reinschmidt, 1996).

Cross talk is primarily a concern for joints that articulate principally about one axis, such as flexion/extension of the knee (Reinschmidt, 1996). Within the context of Cardan angles, not only will tibiofemoral flexion/extension be registered, flexion will be cross talked into ab/adduction and internal/external rotations (the result from ill defined anatomical co-ordinate systems). To illustrate this, a subject purely flexes the knee 30° which roughly corresponds to the amount of knee flexion occurring during the stance phase of running (Reinschmidt, 1996). The uncertainty in defining the anatomical co-ordinate system is 6° for the internal/external rotation and 3° for the ab/adduction position. The resulting cross talk would be 5.6° in ab/adduction and 6.7° in internal/external rotation.

To enable intra-subject comparisons, Reinschmidt used the same standing trials to define both skin and skeletal based anatomical co-ordinate systems (Reinschmidt, 1996; Reinschmidt *et al.*, 1997a). However, comparisons across subjects may be difficult. Differences may be caused by slight differences in defining the anatomical co-ordinate system. This is particularly a concern when describing skeletal marker motion since uncertainties in defining the anatomical co-ordinate system may cause *cross talk*.

Kinetics: Force Plate Analysis

Traditionally, force platforms have been used in biomechanics for quantifying external forces during human gait (Branch *et al.*, 1989; Branch and Hunter, 1990; Winter, 1990). Ground reaction forces are a reflection of forces imparted to the foot by the ground and composed of the sum of multiple forces generated by the body as a system during an event (Branch *et al.*, 1989). A force plate not only yields 3D ground force vector components (vertical load, fore-aft shear, and medial-lateral shear), but also gives torque about the vertical axis, and body centre of pressure location. Changes in these forces may reflect changes in the extremity. It should be noted that the centre of pressure does not give an indication as to how ground reaction forces are distributed under the area of contact.

Impact magnitudes (vertical forces) can be as high as 4000 N (about 5 times body weight) (Winter, 1990). This initial spike of force in synchronisation with axial acceleration of the leg could be responsible for tibial fracture, cartilage damage, and ligament over-stress. During impact or initial ground contact, the foot pushes in the anterior direction and the reaction force from the force plate is directed in the posterior direction. After initial ground contact, the foot pushes in the posterior direction. Consequently, the reaction force from the plate is in the anterior direction. The medio-lateral component often shows an initial reaction force in lateral direction that results from a medial movement of the foot during landing. This initial lateral force is usually shorter than 20% of the total contact time and is usually followed by a reaction force in the medial direction that is often present during the rest of the ground contact time which is usually smaller than the initial lateral force (Nigg, 1994). Medio-lateral variability among intra- and inter-individuals are larger than for vertical and anterior-posterior force time curves (Nigg, 1994).

In examining the effects of functional bracing on ACLD subjects using force platforms, increases in both vertical and antero-posterior ground reaction forces during initial impact provide evidence that bracing alters the kinematics of the lower limb (Cawley *et al.*, 1991; DeVita *et al.*, 1992; Nigg, 1994). Non-braced ACLD individuals generated lower vertical and antero-posterior shear forces than those having braced ACLD knees and non-braced normal knees. Particularly in cutting manoeuvres, braced ACLD knees yielded significantly greater shear forces than non-braced.

Tibone *et al.*, (1986) investigated a group of non-braced ACLD subjects performing a variety of functional activities and compared sagittal shear forces and vertical forces between limbs. During free walking, no significant differences were reported between limbs. However, significant increases were reported in midstance vertical forces and significantly lower toe-off vertical forces for the deficient limb during fast walking and running respectively. It has been speculated the higher midstance vertical force decreased the forces across the joint by "flattening the curve". It appears ACLD subjects compensate during walking and running by attempting to diminish forces by putting less weight on the limb during the plant. During the cross cut and side cut, lateral shear forces were lower for the deficient knee. Antero-posterior shear forces were also lower during the cross cut. Vertical forces were significantly lower for the involved limb during the cross cut only. It seems that ACL deficient individuals use multiple techniques to survive a cut without subluxing the knee. One uses a slower approach to the cut, spends more time in the stance phase (plant) of the cut, reduces the angle of the cut and exerts less force on the planted leg during the cut (Tibone *et al.*, 1986).

Cook *et al.* (1989) investigated ACLD athletes performing cutting and running manoeuvres employing Tibone's (1986) protocol. Subjects performed the functional tests using both limbs although the deficient limb was tested during braced and non-braced conditions. Quadriceps and hamstring torque were recorded using an isokinetic machine and manual displacements were measured with the KT 1000 arthrometer for both limbs post exercise. All non-braced limbs yielded greater displacements. Ground reaction data and kinematic data were recorded between limbs and across subjects during straight running, straight cutting and cross cutting. The straight cut required to plant with the reference limb and cut to the opposite side. For the cross cut, subjects brought the swing limb across the front of the body and cut to the same side as the planted limb.

Comparisons between braced and non-braced conditions for the straight cut and cross cut revealed small increases in ground reaction forces with significant differences in the sagittal plane during the straight cut for the braced ACLD limb (Cook *et al.*, 1989). Normal knees produced significantly greater sagittal forces while the ACLD leg was braced than when non-braced during both cutting manoeuvres. Otherwise, no significant differences were evident in approach times, cutting angles or time on the force plate between limbs or between conditions.

It has been suggested the force changes may be the result of the brace's ability to control damaging forces about the knee and subsequently improve athletic performance. However, the increase in weight of the extremity owing to the brace or the change in confidence level while wearing the brace must be accounted for. Thus, the increases evident in both vertical and fore-aft forces may be attributed to this increase in weight.

The reputed minimum strength allowable for return to sports participation is 90% of the sound limb torque. All but 1 subject achieved a 90% or greater value on *hamstring* torque and this may reflect prior emphasis on hamstring conditioning. However, of the 14 subjects, 5 did not achieve 80% of their sound limb *quadriceps* torque. Of interest are the significant differences that occurred when comparing braced with non-braced limbs during running with weak (< 80% torque of sound limb) vs. normal quadriceps strength. During straight line running, the braced weaker group of athletes produced less lateral and aft forces while simultaneously achieving a faster velocity. This may suggest excessive lateral and aft shear forces evident during ACL deficiency may be detrimental to forward velocity (Cook *et al.*, 1989).

Despite high speed film and skin markers, the cutting angles could not be measured with an error of less than 10^{o} on repeated determinations and perhaps this gross margin of error contributed to the inability to observe statistical differences in the cutting angles during brace wear.

To determine whether functional bracing altered biomechanical parameters during dynamic testing, Valias *et al.* (1990) evaluated normal and ACLD subjects during the performance of a cross cut. Both groups performed three separate bracing conditions; no brace, a placebo and a DonJoy (polycentric) brace. Subjects ran to a marked position on a force platform and cut sharply as fast as possible. The ACLD group also tested their sound leg without any brace to establish normal values for comparison against their deficient limb. Two performance parameters were measured; the speed of approach and the acuteness of the cut (cutting angle). Biomechanical parameters focused on peak vertical force at impact and torque about the cutting extremity. In all test conditions, no significant differences in performance parameters were reported between groups. Therefore, changes in the biomechanical parameters cannot be attributed to performance differences, rather differences can be related to the brace itself.

Valias *et al.* (1990) reported no significant differences in vertical forces during cutting between groups for all conditions. However, differences in torque were evident. When tested on their sound leg, normals produced less torque when wearing the functional brace than in the placebo or non-braced condition. When ACLD subjects were tested on their deficient limb, the functionally braced limb and non-braced limb had lower torque compared to their sound leg but there were no statistical differences between the braced and non-braced involved limbs. A statistically significant decrease in torque was found between limbs when individuals used a functional brace. When wearing the placebo, no significant differences were evident between the placebo and sound leg suggesting that deficient subjects can produce normal stress if they sense the security of a brace. But both normals and ACLD subjects tended to have lower torque with the brace than without it suggesting that functional braces have a biomechanical constraining effect that prevents the leg from generating higher forces than it is capable of doing without a brace.

Clearly though, no study showed that forces at the knee were diminished in a protective fashion by using a derotational brace during a dynamic activity.

One Legged Jump

The one-legged hop (OLH) is a common functional knee evaluation test to assess knee instability after an ACL injury (Gauffin *et al.*, 1990b). Performance scores derived from functional knee tests provide objective measurements of disability related to a specific situation (Lysholm and Gillquist, 1982; Tegner and Lysholm, 1985; Tegner *et al.*, 1986). Their value lies in evaluating dysfunction in various daily and competitive activities after ACL injury. Functional testing has become more prevalent since a strong correlation has been shown to exist between subjective dysfunction and performance during specific conditions (Tegner and Lysholm, 1985; Tegner *et al.*, 1986).

Gauffin *et al.* (1990b) examined the basic function and performance of a 3-point functional knee on unilateral ACLD subjects. Fifteen unilateral ACLD subjects and 12 normal subjects of the same activity level were selected. Using a modified Lysholm Knee Scoring Scale (Lysholm and Gillquist, 1982), ACLD patients subjectively evaluated knee stability when performing common everyday tasks. This functional instability rating scale (scored out of 100) monitors a person's subjective reactions to loss of knee function and symptomology. The more often symptoms arise, the lower the score (Table B-7). Since subjective scores are dependant on activity levels, it is important to relate this score to an activity score. Subsequently, patients completed an Activity Grading Scale (Tegner and Lysholm, 1985) in which certain activities are rated according to how troublesome they are to perform. This functional score is graded from 0-10 listing both daily activities and competitive sports (Table B-8). A score between 5-10 can only be achieved if the patient participates in recreational or competitive sports. Activity levels were subsequently analysed in relation to the Lysholm Knee Scoring Scale (Lysholm and Gillquist, 1982).

Afterwards, subjects performed isokinetic flexion/extension tests to determine peak muscle torque at 0^0, 30^0, & 180^0/s. The testing protocol required subjects perform the following standard knee performance tests:

1. Running 2 laps of a figure-eight course for a total of 40 m. Total time and separate curve times were recorded using photoelectric cells.

2. Prior to testing, three jumps for maximal horizontal distance and landings for each leg were performed with hands behind the back. The longest measurement for each leg was marked on the floor to determine proper take off distance to the force platform. Testing required subjects jump from this mark to the force platform without especially trying to hit it. The first leg tested was randomised then the same procedure was repeated for the contralateral leg. All tests, except muscle strength test, were carried out with and without the brace. Jump distance ratio between the injured and non-injured leg was calculated.

Since ACLD patients participated in a strength training program prior to testing, activity levels were scored at 7 which corresponds to recreational sports (Tegner and Lysholm, 1985; Gauffin *et al.,* 1990b). Although isokinetic testing revealed symmetrical peak torque's between the injured leg and non-injured leg, most patients reported chronic instability during athletic activity. The mean Lysholm Knee Score for the ACLD group was 87 ± 12 and ranged between 52 to 100.

During tests that greatly stressed the knee joint, impaired performances were evident. Jump distances were significantly longer for the control group and the ACLD groups normal limb compared with their deficient limb. No correlations were reported with reduced muscle strength. Significant differences were also reported in curve times between the patient group and the control group for the figure eight run. However, the brace did not affect the ACLD group's total run time or jump distances. Gauffin *et al.*, (1990b) reported no significant differences between brace and non-braced conditions. Bracing did not improve jump distances or reduce the total time to run the figure eight course suggesting the brace does not significantly alter performance.

Gauffin *et al.,* (1990a) examined both the kinematic and kinetic parameters of the OLH at the moment of landing. Sixteen ACLD subjects with complete ACL rupture were evaluated. The Selspot motion analysis system integrated with a force plate recorded the kinematics and kinetics respectively. LED's were placed over the anterior-superior iliac spine, the greater trochanter, the axis of rotation of the knee, the lateral malleolus, and the head of the fifth metatarsal to define four body segment links. Surface electrodes placed over the rectus femoris and long head of the biceps femoris recorded neuromuscular activity simultaneously at the moment of landing (Gauffin *et al.,* 1990a).

Prior to testing, subjects performed isokinetic flexion-extension tests as outlined above. Gauffin *et al.* (1990a) reported ACLD subjects attained symmetrical peak torque or exhibited a mean 4-7% deficit in thigh muscle strength for the injured limb at 30^0/s and 180^0/s during knee extension and at 30^0/s for flexion compared to the contralateral leg. Although total laxity was greater for the injured leg, the Lysholm Knee Score as well as Activity Grading Scale determined that ACLD subjects had excellent/good knee function. Although subjects participated in recreational sports and were considered rehabilitated, patients still exhibited impaired performances in functional tests when rehabilitated to this level.

Subjects performed the OLH as described earlier. The first leg tested was randomised and was repeated for the contralateral leg. Knee angles and angular velocity was calculated in the sagittal plane and related to their peak vertical force at heelstrike.

Jump distances were significantly shorter for the deficient limbs compared to their non-injured legs and the reference group as reported in Gauffin's (1990a) study. Interestingly, non-injured jump distances were shorter compared to the previous reference group. Upon footstrike onto the force platform, coinciding events of valgus thrust to the foot, eccentric loading in the knee and quadriceps activity were noted. At peak loading, the ground reaction force (GRF) and knee joint angle followed a consistent pattern:

1. There was a *Valgus* thrust represented by a laterally directed shear force on the foot (Fy).
2. Peak loading coincided with initiation of eccentric flexion of the extended knee and there was a tendency for the injured knee to be somewhat more flexed than the non-injured.
3. Knee angular velocity was negative (an eccentric motion during deceleration) at peak loading and this was significantly lower for the injured limb compared to the normal leg.

No significant differences were evident in knee flexion angles at impact although significant lower eccentric angular velocities for the injured leg were recorded. Ground reaction forces showed no significant differences between injured and non-injured limbs and the angle of the sagittal ground reaction force remained similar between the affected and non-affected limbs.

Myoelectric patterns for the two muscle groups were consistent for both limbs with a peak hamstring activity close to touch down and peak quadriceps activity 100-200 ms later. The peak quadriceps activity coincided with the peak negative angular activity. Gauffin *et al.*, (1990a) purported that any one or combination of these events may have contributed to knee instability. This impairment in performance (shorter jump distances) may depend upon reduced muscle strength, the adaptation of motor control, or restraint caused by the fear of possible subluxation at impact.

Gauffin *et al.*, (1990a) suspected ACLD subjects alter their motion patterns similar to those found during side-step cutting manoeuvres. Adaptation may result from the lower negative angular velocity. The correlation between peak knee flexion torque at 180^0 /s and peak angular velocity at loading might be interpreted as added hamstring function resulting in a compensatory mechanism to diminish knee subluxaion. Branch *et al.* (1989) during cutting showed an increase in electromyographic (EMG) activity in the hamstrings and decreased in the quadriceps during the stance phase compared to normals. By having the knee in a more flexed position, the hamstrings are in a better position to prevent excessive anterior translations and internal/ external rotations (Gauffin *et al.*, 1990a). An increase in hamstring activity working synergistically with the ACL combined with reduced antagonistic quadriceps activity could enhance this effect.

Gauffin *et al.*, (1992) investigated whether rehabilitated ACLD subjects alter motor control to reduce joint instability when performing the OLH (adaptations resulting from a measurable decrease of sagittal shear force). Nine ACLD subjects exhibiting unilateral ACL rupture were selected for testing. Significant lower peak extensor muscle torque was reported for the injured leg both at 30^0 and 180^0 respectively and for knee flexion at 180^0 /s. Total laxity was greater for the injured leg. However, six subjects scored high on the Lysholm Knee scale and three reported chronic instability during athletic activity. The median activity level was six and ranged between four (moderate heavy work) and nine (competitive) activity.

For intra- and intersubject comparisons, subjects performed a seated flexion and extension MVC with the knee flexed 60°. EMG's were expressed as percentage of the maximal voluntary contraction (MVC). Gauffin *et al* (1992) reported differences in movement patterns and EMG activity between the injured and non-injured limb. Lower quadriceps activity was reported for the ACLD limb at footstrike but no differences in hamstring activity. Branch *et al.* (1989) has shown increased hamstring activity and decreased quadriceps activity during stance for ACL deficient knees when performing side-step cuts. Gauffin's *et al.*, (1992) findings can be interpreted as having a protective effect on the knee joint.

Greater hip and knee flexion angles were observed upon footstrike with greater knee flexion angles at peak vertical force. The injured limb showed a significant reduction in quadriceps activity along with significantly lower peak external torque and peak knee angular velocity but greater maximal knee flexion angles. Increased in hip flexion angles place the hamstrings in better position to prevent pathological translations and rotations (Gauffin and Tropp, 1992). Renström using cadavers knees has shown that quadriceps activity significantly increased the strain within the ACL and that with greater flexion angles, coactivation of hamstrings during knee extension can reduce strain (Renstrom *et al.,* 1986).

According to Gauffin *et al.,* (1992), the differences in test performances may not directly result from ACL deficiency, rather from altered movement patterns in compensating for the injury. Branch *et al.* (1990) has shown an increase in hamstring activity with a concomitant reduction in quadriceps during stance when performing side-step step cutting manoeuvres. This could have a protective effect on a deficient knee while affecting performance. Although Gauffin's subjects were well rehabilitated, there were minor deficits in muscle strength which may be a possible cause for differences in measurements.

Methodology

Subjects

Six anterior cruciate deficient (ACLD) subjects having no prior surgical treatment will be selected by an orthopaedic surgeon from the Sports Medicine Institute located at the Karolinska Hospital in Stockholm Sweden. Subjects will have a history of significant instability exhibited by frequent episodes of giving way, causing them to modify their activity level. Patient's knees will exhibit at least a +3 laxity score compared to their contralateral leg when clinically evaluated using the KT 1000 arthrometer. All subjects will have signed an informed consent form and a medical release form in accordance with the Karolinska Institute.

Prior to the pin implantation, patients will complete the Lysholm Knee Scoring Scale (Lysholm and Gillquist, 1982) to assess their loss of knee function (Table B-7). This is a discrete rating scale scored out of 100 to evaluate the patients' symptoms during the performance of daily activities. An Activity Grading Scale (Tegner and Lysholm, 1985) will also be completed which rates certain activities according to how troublesome they are to perform (Table B-8). This functional score is graded from 0-10 listing both daily activities and competitive sports. A score between 5-10 can only be achieved if the patient participates in recreational or competitive sports. Their activity levels will be later analysed in relation to the Lysholm Knee Scoring Scale.

Surgical procedure

Intracortical Steinmann bone pins (2.5 mm diameter) will be inserted with a manual orthopaedic drill into the subjects deficient leg. Unlike previous studies (Levens *et al.*, 1948; Lafortune *et al.*, 1992; Lafortune *et al.*, 1994; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b), the knee is flexed 45° prior to pin implantation in order to minimise impingement problems with the iliotibial band (McClay, 1990). The pins will be inserted anterolaterally and superior to the femoral condyle and antrolaterally in the proximal portion of the tibia. This insertion site will ensure that no impingement between the brace and pin/target markers occur during the dynamic functional task (McClay, 1990). Prior to insertion, the skin, subcutaneous tissue and perioteum are anaesthetised with standard anaesthetic. The anaesthetic is generally active for 2 hours leaving ample time for the motion recordings. Target markers will then be affixed to the pins. Each triad is comprised of four noncollinear 7 mm reflective markers, one in the centre and three attached to orthogonal projecting rods as seen in Figure A-3. The intracortical pins will remain inserted during the single test session.

Once the pins are implanted, stereo-photogrammetric x-rays will be taken in order to record the position of the markers and to define the tibial and femoral anatomical reference points. The deepest point of the intercondylar groove is defined as the origin for the femoral co-ordinate system. The longitudinal axis passes through the origin and is directed superiorly and parallel to the long axis of the femur. The medio-lateral axis progresses along a line connecting the most distal points on the medial and lateral femoral condyles, passes through the origin and is perpendicular to the longitudinal axis. The remaining axis is calculated using the cross product of the two defined unit vectors. The origin for the tibial co-ordinate system is located on the most proximal point of the medial intercondylar eminence. A line parallel to the longitudinal axis of the tibia passes through the origin and is used to define the proximal-distal axis. The medio-lateral axis progresses along a line through the estimated centres of the medial and lateral tibial articular surfaces passing through the origin and is perpendicular to the longitudinal axis. The remaining anterior-posterior axis is calculated using the cross product (Lafortune, 1984; McClay, 1990; Lafortune *et al.*, 1992; Reinschmidt, 1996; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b)

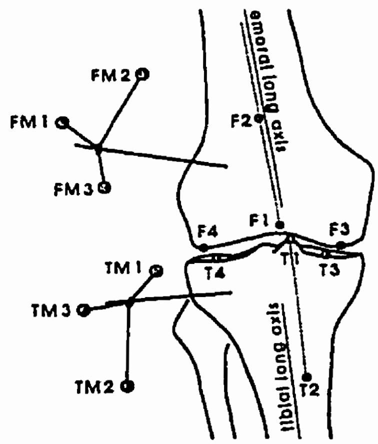

Figure A- 3: RSA picture outlining the points digitised to establish the tibial and femoral anatomical co-ordinate system.

(Reinschmidt, 1996)

Motion recordings

Six infrared 60 Hz MacReflex cameras will be paired and affixed to specially designed tripods to record the motion. The MacReflex motion analysis system will be configured so that the two cameras in each pair record at alternate frame sequences. This is equivalent to three twin cameras sampling at 120 Hz. Each camera is equipped with f12.5 lenses to give a narrow horizontal field of view of 28° thereby zooming in on the knee. When the camera positions are set with respect to the calibration frame, each camera view is verified to determine that the calibration frame markers are satisfactorily seen from each camera view. Prior to recording, the measurement area approximately 45 cm off the floor (representative of knee height) will be calibrated using calibration frame equipped with nine control points (volume 25 x 49 x 15 cm^3). Camera pairs will be orientated to obtain a field of view covering the entire measurement area (Figure B-2). A *Merit Value* under 5.0 indicates a good calibration set-up. After MacReflex calibration, the orientation of the laboratory system used by the motion analysis system is known (Figure B-1). All target markers will be visible in all cameras throughout the loading and stance phase during motion recordings.

Following calibrations, a standing reference trial must be recorded with the subject in a controlled posture. To record the reference file, the subject is aligned so that their sagittal plane is oriented with the x-z plane of the MacReflex-calibrated system (with the z-axis is directed vertically). While the subject remains motionless in an upright standing position, this position is recorded for the reference sequence (Figure A-4). Two additional 7 mm markers are placed on the corners of the force platform to set the correct aperture for the MacReflex cameras for each motion recording.

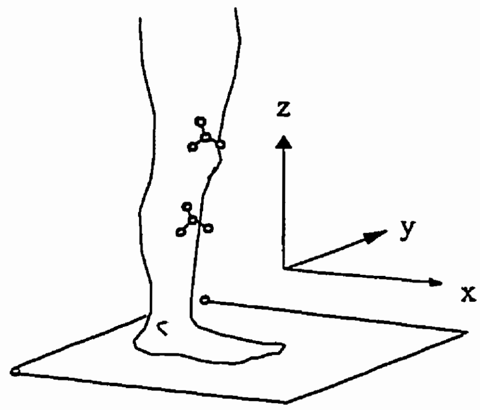

Figure A- 4: Triad markers attached to the thigh, lower leg and reflective markers placed at the corners of the force plate.

(adapted from Karlsson 1997)

Force Plate Recordings

Both Macreflex and a Kistler force platform will be synchronised to record simultaneously via an external trigger for a collection time of five seconds. The force platform will record ground reaction forces at a sampling rate of 960 Hz. Peak vertical forces (Fy) and anterior-posterior shear forces (Fx) will be analysed. Kinematic data will be recorded via Macreflex whereas Kinetic data collected with Zoom. All data will be converted from analogue to digital through the MacReflex and Zoom A/D board. Kinematic and kinetic data will be converted to ASCII format and then saved in the memory of a personal computer for later analysis.

Knee Brace

The DonJoy Legend knee brace will be selected by the researcher and applied according to the specifications prescribed by the manufacturer. To determine the correct brace size, specific measurements must be taken prior to the surgery. The calf is measured from the mid-point or widest circumference and the value recorded. Thigh measurements are taken 15 cm from the superior aspect of the patella and the diameter recorded from this point. The following Table A-1 lists brace sizes according to leg dimensions. Once the dimensions are determined, the appropriate brace is selected and fitted to the subject.

Calf cm	Thigh cm
35-36 Sm.	40-47 Sm.
37-38 M.	48-53 M.
39-40 Lg.	54-60 Lg.
	>60 XL.

Table A- 1: Sizing chart for the DonJoy Legend Functional Knee brace.

Brace application must follow the correct strap sequence beginning with No. 1 to help position the brace on the leg. The centre of the hinge is lined up at or above the top of the kneecap and resting slightly back of the midline of the leg as illustrated in Figure A-5. Misalignment could lead to discomfort and create alterations in moments and forces at the hinge. The uniaxial polycentric hinge is set at the 10° extension stop which is the generally accepted standard to prevent hyperextension of the knee (DeVita *et al.,* 1992). The brace is coloured black to reduce reflections in the MacReflex recordings and enable greater accuracy in marker identification.

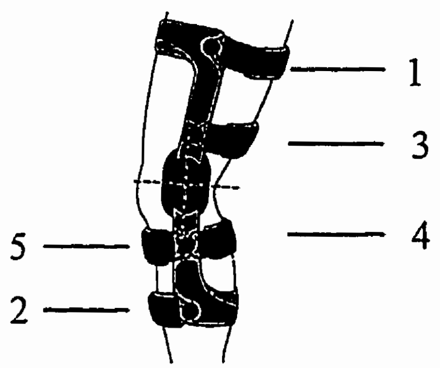

Figure A- 5: DonJoy Legend ACL Brace indicating proper fitting sequence

Experimental protocol and set-up

After pin insertion, each subject is given ample time to perform the *One Legged Hop* (OLH) to familiarise themselves with the pins and testing protocol. To sufficiently stress the ACL, each subject will maximally hop for horizontal distance. From an initial standing position with the deficient limb set back, the subject pushes off from their sound limb and lands on their deficient limb. Their longest measurement is recorded and marked on the floor to determine the proper take off distance to the force platform. Testing required subjects jump from this mark to the force platform without especially trying to hit it. After familiarisation with the procedure, two standing reference trials and five measurement trials are recorded. Standing reference trials are recorded prior to and following the measurement recordings. For the standing trial, subjects stand in a neutral position and align their feet parallel to the force platform in order to define the tibial and femoral anatomical co-ordinate system. It is assumed the segmental co-ordinate systems are aligned with the global co-ordinate system during standing.

Each subject is to be tested during a single experiment session, wearing their own running shoes and dark lightweight clothing for ease in identifying markers. Subjects will be randomly assigned to start with either the braced or non-braced condition. After the standing trials and five measurement trials are completed for the first test condition, two additional standing trials and five measurement trials will be collected for the subsequent testing condition. Synchronisation between the jump and data collection will be initiated with via a verbal que. Having given the command to start, data collection and the performance of the jump will commence.

Data reduction and analysis

Three-dimensional reconstruction

From each camera pair, both the standing and measurement trials for each subject will be manually sorted and autotracked using MacReflex 3.2 PPC data acquisition software. The 2-D image co-ordinates from each camera are digitised into Cartesian spatial co-ordinates and transformed onto a 3D co-ordinate system employing MacReflex's Direct Linear Transform (DLT) algorithms. MacReflex's autotracking calculates the 3D spatial reconstruction for each successive frame. All cameras will be used for the three-dimensional reconstruction. Any incorrect markers will be invalidated and if invalid markers or drop-outs are evident, these will be filled using linear interpolation (interpolations must be done between actual marker appearances).

After autotracking, the data is exported in the TSV format (text files) including the frame numbers and saved as the reference file *ref.TSV* or motion *motion.TSV*. The frame numbers (included in the TSV-files) are kept for the *Segment Analysis* (©Karlsson, 1997) calculations so that the frames of the output files will correspond to the numbers of the original MacReflex recording.

Reference frames and relative orientation

Three-dimensional skeletal motion will be derived using specially written software (*Segment Analysis*: The Lundberg Laboratory for Motion Analysis Göteborg) employing the frames of reference (Figure A-6) and algorithms described by Lafortune *et al.*, (1992). The following descriptions are from the Segment Analysis instructional manual (© Karlsson, 1997) unless otherwise stated.

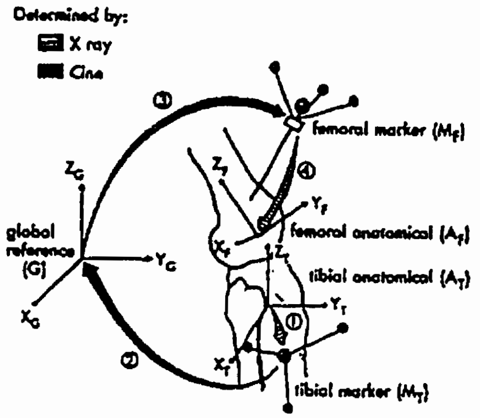

Figure A- 6: Illustration of Lafortune's frame of reference
(Lafortune *et al.*, 1992)

The software allows for the analysis of the relative motions between segments moving in space. Analysis includes relative 3-D motions of the triad markers attached to both the measurement segment and reference segment as well as two fictive points for each respective segment. The results are expressed as either relative 3-D angular orientation or relative 3-D displacements between the two fictive points. Prior to running the segment analysis program, a reference position and successful motion recordings should have been made.

The anatomical co-ordinate system utilises Grood and Suntay's "joint co-ordinate" system and is normalised with respect to the neutral standing trial (Grood and Suntay, 1983). During neutral standing, the segmental (anatomical) co-ordinate systems are assumed to be aligned with the global co-ordinate system. To describe movement of the segmental (anatomical) co-ordinate systems, three rotational and three translational degrees of freedom are employed resolving the femoral anatomical co-ordinate system into the tibial anatomical co-ordinate system and normalised to the standing reference trial. These calculations are derived from the bone markers and the methods used to calculate the transformation matricies (Lenox and Cuzzi, 1978) are reported in greater detail elsewhere (Lafortune, 1984; Lafortune *et al.*, 1992).

General joint motion is partitioned into 6 familiar anatomic motions and is based on Cardan or Euler angles. According to the conventions described by Grood and Suntay (1983), flexion/extension and medial-lateral shift occur around the fixed medio/lateral femoral axis, ab/adduction and anterior-posterior drawer around the floating axis and internal/external knee rotation around fixed tibia proximal/distal axis.

The kinematic ASCII data derived from the Segment Analysis software will be imported into Bioproc and cut-off frequencies will be determined by running a Fourrier analysis of the angular and translational data respectively. The co-ordinates will then be filtered with a Butterworth 4th order, low-pass, critically damped, zero-lag filter. Additionally, jumps will be time normalised to 100% for each subject and condition and an ensemble average derived. Data reduction will focus solely on the stance phase with emphasis on determining whether differences are evident between conditions and subjects. The following parameters will be reported:

- Absolute 3D angular data at HS and Peak Flexion
- Absolute 3D linear data at HS and Peak Flexion.
- Range of Motion for tibial translation following HS
- The shape and magnitude of the angular curves after normalisation.
- The shape and magnitude of the linear curves after normalisation.

Force plate recordings

Peak vertical load and anterior posterior shear forces will be derived from the force platform data. Ground reaction forces (GRF) will be scaled to body weight and interpolated so that each body position during the landing will have a corresponding applied ground reaction force. Initial contact with the force platform will be noted to co-ordinate film and GRF data. Analysis will focus solely on peak vertical force and anterior posterior shear force. If peak vertical forces are similar for both jumping conditions, any differences in translational data may be attributed to the brace rather than differences in jumping. Jumps will be time normalised to 100% for each subject and condition using the same time criteria established for the MacReflex (kinematic) data. Data reduction will focus solely on the stance phase with emphasis on determining whether differences are evident between conditions and subjects.

Statistical Analysis

Descriptive analysis and descriptive statistics will be performed owing to the small population size. To examine the effect of bracing vs. non-bracing, mean kinematic values (absolute angular and linear data) and standard deviations across trials and conditions will be calculated. Additional kinematic parameters will be investigated: knee position at upon contact with the force platform; maximum knee flexion, peak vertical force, anterior/posterior and valgus/varus ground reaction forces.

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Appendix B

Tables

Table B- 1: Summary of angular data for the tibiofemoral joint during walking

Walking	Flexion/extension	Abduction/adduction	Internal/external Rotation
Leven's			Internal rotation of 3.5° from late swing to MS. Beyond full weight bearing, slight external rotation of 1.5° followed by 0.5° internal rotation. External rotation of 3.5° at TO.
Lafortune	0° - 10° flexion at HS. 15° - 20° flexion to 20% stance. ≅ full extension at 60% stance. Flexion following MS to TO. ROM 40°.	Tibia remained abducted 1.2° throughout stance. Little or no ab/adduction movements.	Two internal rotations evident: 5° from HS to 25% stance. 5° at 70% stance.
Reinschmidt	0°- 10° flexion at HS. 15°- 20° flexion to 20% stance. ≅ full extension at 60% stance. Flexion following MS to TO. ROM 40°. Skin and skeletal patterns similar in shape (mean difference 2.1°)	No general patterns of ab/adduction. Greater ROM (5° - 10°). Dissimilar shape in skin and skeletal ab/adduction curves (mean difference 2.4°)	Subjects either initially internally or externally rotated. Internal (2° to 5°) External 4° ROM from 5° to > 10°. Skin and skeletal int/ext rotation curves dissimilar (mean difference 3.9°)

Table B- 2: Summary of linear data for the tibiofemoral joint during walking

Walking	Anterior/posterior	Medial/lateral	Distraction/Compression
Lafortune	Pattern similar to flex/ext curve Tibia drawn posterior during flexion. 3.6 mm to 50% stance. Tibia drawn anterior during extension. 1.3 mm past neutral (0°).	Patterns similar to flex/ext. 2.3 mm medial shift in flexion 1.5 mm lateral shift in extension.	Following HS, a 3.2 mm distraction during flexion before returning to the zero position as the knee reached maximal stance extension. 0.2 mm compression with extension.

Table B- 3: Summary of angular data for the tibiofemoral joint during running

Walking	Flexion/extension	Abduction/adduction	Internal/external rotations
Lafortune (walking)	0° - 10° flexion at HS. 15° - 20° flexion to 20% stance. ≅ full extension at 60% stance. Flexion following MS to TO. ROM 40°.	During stance, little or no ab/adduction movements. Tibia remained abducted 1.2° during stance.	Two internal rotations evident: 5° from HS to 25% stance. 5° at 70% stance.
McClay	Normals / PFP Angle at HS: -13.83° / -6.76° Peak Flex: -33.96° / -30.59° Peak Ext: -13.33° / -4.7° Angle at TO: -18.99° / -7.09° ROM: 21° / 26° (*positive indicates extension.) 10° - 20° flexion at HS. 30° - 40° flexion to 40% stance. ≅ full extension prior to TO stance. Flexion followed MS to TO. Greater flexion angles for normals throughout stance	Normals / PFP Angle at HS: -6.98° / -0.51° Peak Add: -1.91° / +6.24° Peak Abd: -9.41° / -1.93° Angle at TO: -9.41° / -1.85° ROM: 7.51° / 8.17° (*positive indicates adduction.) Clear ab/adduction patterns Normals exhibited an initial 6° tibial adduction from HS to 40% MS upon which the tibia adducted 8° until TO. Ab/adduction curves similar in shape across subjects and groups. Peak amplitudes higher for patellofemoral pain group (PFP). ROM 8° for both groups	Normals / PFP Angle at HS: 3.81° / 0.23° Peak int: 14.52° / 12.67° Peak ext: 2.81° / 0.88° Angle at TO: 4.38° / 2.52° ROM: 11.71° / 12.44° (*positive indicates int. rotation) Similar patterns between subjects and groups with slight differences in magnitude. Initial internal rotations until MS then external rotations. ROM varied slightly between groups
Reinschmidt	Skeletal flexion/extension curves similar in shape and magnitude across subjects and corresponded well with McClay's investigation. Small differences in knee position at HS were evident across subjects ranging from 0° to 15°. Skeletal and skin based Flex/ext curves were similar in shape and magnitude (relative difference 5°).	No general ab/adduction patterns. Subjects initially either adducted (4°) or abducted (6°- 9°). Poor agreement between skin and skeletal ab/adduction patterns (mean difference 4°)	Similar patterns across subjects with slight differences in magnitude. Compared favourably with McClay. (based on neutral standing trial) From HS to MS, small or pronounced internal rotations were evident varying from 2° to 7° - 9° and followed by external rotations. Poor agreement between skin and skeletal patterns. Reinschmidt's RSA and neutral standing trials agreed well but offset evident between curves

Table B- 4: Summary of linear data for the tibiofemoral joint during running

Walking	Anterior/posterior	Medial/lateral	Distraction/Compression
Lafortune (walk study)	Pattern similar to flex/ext curve Tibia drawn posterior during flexion. 3.6 mm to 50% stance. Tibia drawn anterior during extension. 1.3 mm past neutral.	Patterns similar to flex/ext. 2.3 mm medial shift in flexion 1.5 mm lateral shift in extension.	Following HS, a 3.2 mm distraction during flexion before returning to the zero position as the knee reached maximal stance extension. 0.2 mm compression with extension.
McClay	Normal PFP Posn. at HS -13.31mm -4.2mm Peak post. -17.23mm -7.03mm Peak ant. -10.91mm +3.12mm Posn. at TO -15.77mm -2.56mm ROM 6.32mm 10.15mm (*positive indicates ant. transln.) Patterns similar in shape across subjects and corresponded well with Lafortune's data although running yields higher magnitudes. At HS, the tibia is posterior with respect the femoral origin and is further drawn posteriorly to 25% - 35% stance after which it moves anteriorly. The PFP group had a more anterior position of the tibial reference point & exhibited greater translations than normals.	Normal PFP Posn. at HS -1.88mm -0.79mm Peak med. -10.73mm -8.23mm Peak lat. -0.94mm -2.11mm Posn. at TO -0.94mm -2.17mm ROM 9.81mm 7.20mm (*positive indicates lateral.) A general patters of medial displacement until 50% stance after which lateral translations occurred until TO	Normal PFP Posn. at HS -20.29mm -8.94mm Peak dist. -24.71mm -14.82mm Peak comp. -20.29mm -11.53mm Posn. at TO -24.01mm -2.90mm ROM 4.43mm 5.89mm (*positive indicates compression.) Distraction during flexion (until MS) followed by compression during extension.
Reinschmidt	Similar to patterns as McClay but magnitudes smaller. 4 mm posterior displacement until MS. Anterior tibial displacement to 80% stance followed by a fast posterior displacement towards end of stance.	RSA patterns opposite to motion as McCaly reported. 3.6 mm lateral shift after 15% stance. From 40% - 80% stance, 4.1 mm lateral shift followed by 1.8 mm lateral shift during last 20% of stance.	RSA patterns similar to McClay 5.6 mm distraction from 10% - 40% stance. 6.8 mm compression 2.8 mm distraction during final 20% stance

Table B- 5: Angular patellofemoral data

Walking	Flexion/extension	Abduction/adduction	Internal/external
Lafortune	From HS - MS, the patella initially positioned 11.8 ° of extension and flexed to a neutral position (0°). At peak tibiofemoral extension, it extended to 12.5° after which it moved to -16.1° until TO during flexion.	Ab/adduction highly variable across subjects. The patella was neutrally aligned (adducted 0.6°) at HS and adducted 2° until approximately 50% stance. Following MS, it abducted to 6.2° until TO.	The patellae remained externally rotated throughout stance. At HS, the patella was positioned about 5° of external rotation and was followed by internal rotation until about 50% stance. Until max extension, the patellofemoral joint externally rotated reaching 8.7° followed by internal rotation to 6° at TO.
McClay	Normals PFP Angle at HS -0.64 ° +3.52° Peak Flex. -12.0° -8.93° Peak Ext. -2.00° +4.07° Angle at TO -7.13° +1.58° ROM 11.3° 13° (*positive indicates extension) Patellofemoral flex/ext similar to tibiofemoral motion but variable across subjects. From HS - 40% stance, the patella was neutrally positioned and flexed to -12°. Extension followed until 90% stance peaking at -2°. The remaining 10% saw the patella flex to -7° at TO. The PFP group exhibited less flexion at HS and less peak flexion at MS but greater ROM compared to normals.	Normals PFP Angle at HS -1.68° -2.49° Peak Abd -5.99° -3.13° Peak Add -1.56° -0.77° Angle at TO -4.78° -3.13° ROM 3.43° 3.22° (*positive indicates adduction) Similar ab/adduction patterns across subjects and groups. From HS – MS, the patella initially abducted followed adduction until peak extension. An additional abductory phase occurred preceding TO. No clear differences between groups.	Normals PFP Angle at HS -1.48° -2.19° Peak Int. Rotn. -0.36° -2.19° Peak Ext. Rotn. -3.87° -6.87° Angle at TO -4.61° -6.87° ROM 3.68° 4.74° (*positive indicates int. rotation) Rotational patterns similar across subjects and groups although offsets resulted in the patella being positioned either in external rotation, in internal rotation or neutrally aligned at HS. Following HS, the patella externally rotated and remained in this position until 50%-75% stance when further external rotation occurred. In general, subjects exhibited very little rotational movements.

Table B- 6: Linear patellofemoral data

Walking	Anterior/posterior	Medial/lateral	Distraction/Compression
Lafortune	The position of the patella at HS was positioned 42.2 mm anterior to the femoral anatomical origin and remained forward throughout stance. Following HS, it briefly displaced posteriorly then gradually moved forward 1.8 mm until maximal tibiofemoral extension. It migrated back towards the femur as the tibiofemoral joint began flexing as TO approached.	At HS, the patellar was 10.8 mm medial with respect to the femoral anatomical reference. Throughout stance, it shifted 7.2 mm laterally until TO remaining 3.6 mm medial to the femur.	Upon HS, the patella continued to move proximally from its initial 27.5 mm proximal position to the femur. After, patterns closely matched tibiofemoral flexion/extension. When the knee flexed, the patella displaced laterally, posteriorly and distally. The patella reached a minimum position of 23.1 mm above the femoral origin at MS . During tibiofemoral extension, it's proximal position peaked at 32.6 mm then moved distally reaching a minimum value of 0.6 mm proximal to the femur at TO.
McClay	Normals PFP Posn at HS 31.03mm 41.66mm Peak ant. 25.11mm 39.48mm Peak pos. 32.92mm 43.89mm Posn at TO 31.61mm 43.38mm ROM 7.82mm 4.44 mm (*positive anterior translation) At HS, the patella was positioned anterior to the femur remaining stationary about 50 ms. During tibiofemoral flexion, the patella displaced posteriorly followed by an anterior movement during extension. The PFP group exhibited little or no posterior movement following heelstrike (1.18 mm) compared to the normal group (5.92 mm). Total range of motion for the NL group was almost twice that of the PFP group respectively.	Normals PFP Posn at HS -9.44mm -5.22mm Peak ant. -11.57mm -12.44mm Peak pos. -9.09mm -5.53mm Posn at TO -10.77mm -11.20mm ROM 2.52mm 6.93mm (*positive lateral translation) General initial medial translations with respect to the femur followed by a lateral translation. In general, the patellae was positioned medial to the femur at heelstrike and remained throughout stance. PFP group exhibited medial-lateral excursions approximately 2.75 times than that of the NL group (6.93 vs. 2.52 mm).	Normals PFP Posn at HS 13.47mm 17.25mm Peak dist. 1.65mm 3.39mm Peak comp. 13.69mm 20.37mm Post at TO 8.37mm 6.93mm ROM 12.12mm 16.99mm (*positive lateral translation) Translations resembled tibiofemoral and patellofemoral flexion/extension patterns. Patterns similar in shape and magnitude across subjects. Generally, the patellar origin remained proximal to the femoral origin throughout support. It distally translated at HS peaking 40% into stance. It then moved proximally peaking at 90% of support followed by a distal translation through toe-off.

Table B- 7: Lysholm Knee Scoring Scale

(Lysholm 1982)

Limp (5 points)

None	5
Slight or Periodical	3
Severe and Constant	0

Support (5 points)

None	5
Stick or crutch	2
Weight bearing impossible	0

Locking (15 points)

No locking and no catching sensations	15
Catching sensation but no locking	10
Locking	
Occasionally	6
Frequently	2
Locked joint on examination	0

Instability (25 points)

never giving way	25
rarely during athletics or severe exertion	20
frequently during athletics or other severe exertion (or incapable of participation)	15
Occasionally in daily activities	10
often in daily activities	5
Every step	0

Pain (25 points)

none	25
Inconstant and slight during severe exertion	20
marked during severe exertion	15
marked on or after walking more than 2 km	10
marked on or after walking more less than 2 km	5
constant	0

Swelling (10 points)

None	10
On severe exertion	6
On ordinary exertion	2
Constant	0

Stair Climbing (10 points)

No problems	10
Slight impaired	6
One step at a time	2
Impossible	0

Squatting (5 points)

No problems	5
Slightly impaired	4
Not beyond 90°	2
Impossible	0

Table B- 8: Activity Score

(Tegner 1985)

10. Competitive sports
Soccer - national and international elite

9. Competitive Sports
Soccer, lower division
Ice Hockey
Wrestling
Gymnastics

8. Competitive sports
Bandy
Squash or badminton
Athletics (jumping etc.)
Downhill skiing

7. Competitive sports
Tennis
Athletics (running etc.)
Motorcross, speedway
Handball
Basketball

Recreational
Soccer
Bandy or ice hockey
Squash
Athletics (jumping etc.)
Cross country both recreational & competitive

6. Recreational sports
Tennis or badminton
Handball
Downhill skiing
Jogging at least 5 times per week

5. Work
Heavy labour (e.g. building forestry)
Competitive sports
Cycling
Cross country skiing
Recreational sports
Jogging on uneven ground at least twice weekly

4. Work
Moderate heavy labour (truck driving, heavy domestic work)
Recreational
Cycling
Cross country skiing
Jogging on even ground at least twice weekly

3. Work
Light labour (nursing)
Competitive and recreational sports
Swimming
Walking in forest possible

2. Work
Light labour
Walking on uneven ground possible but impossible in forest

1. Work
Sedentary work
Walking on even ground possible

0. Sick leave or disability pension because of knee problems

Figures

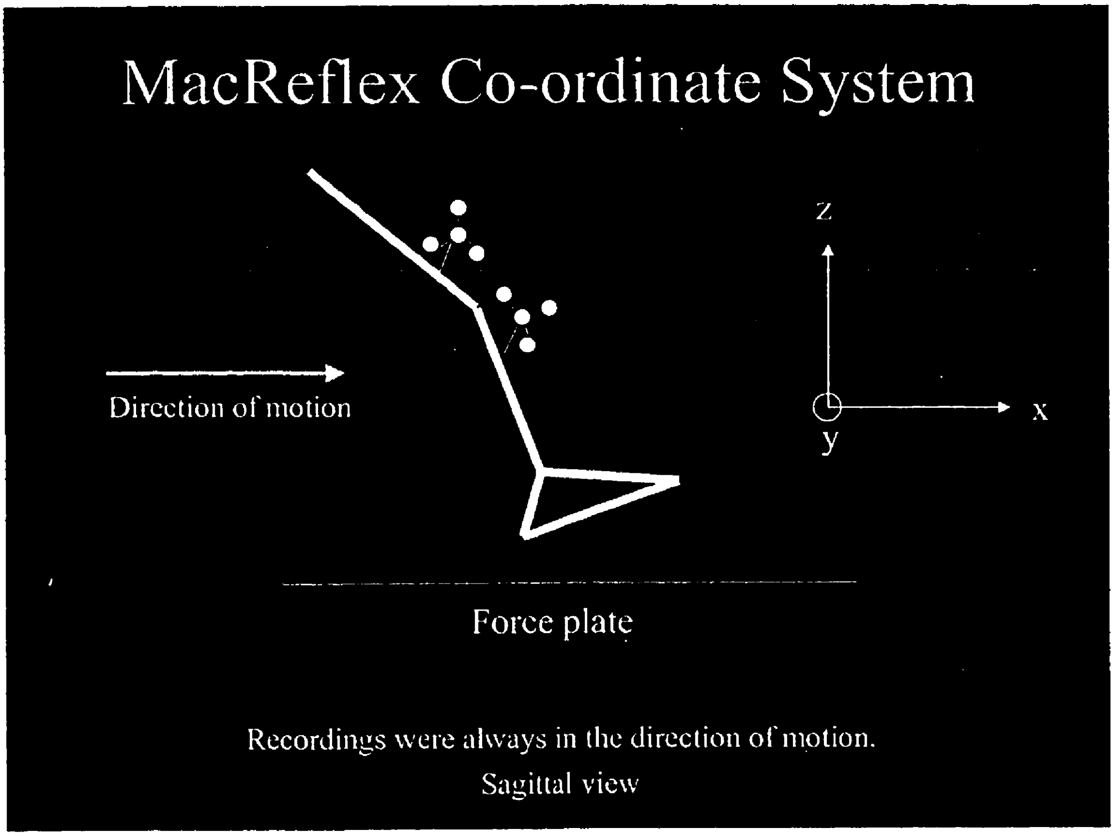


Figure B- 1: Global co-ordinate system

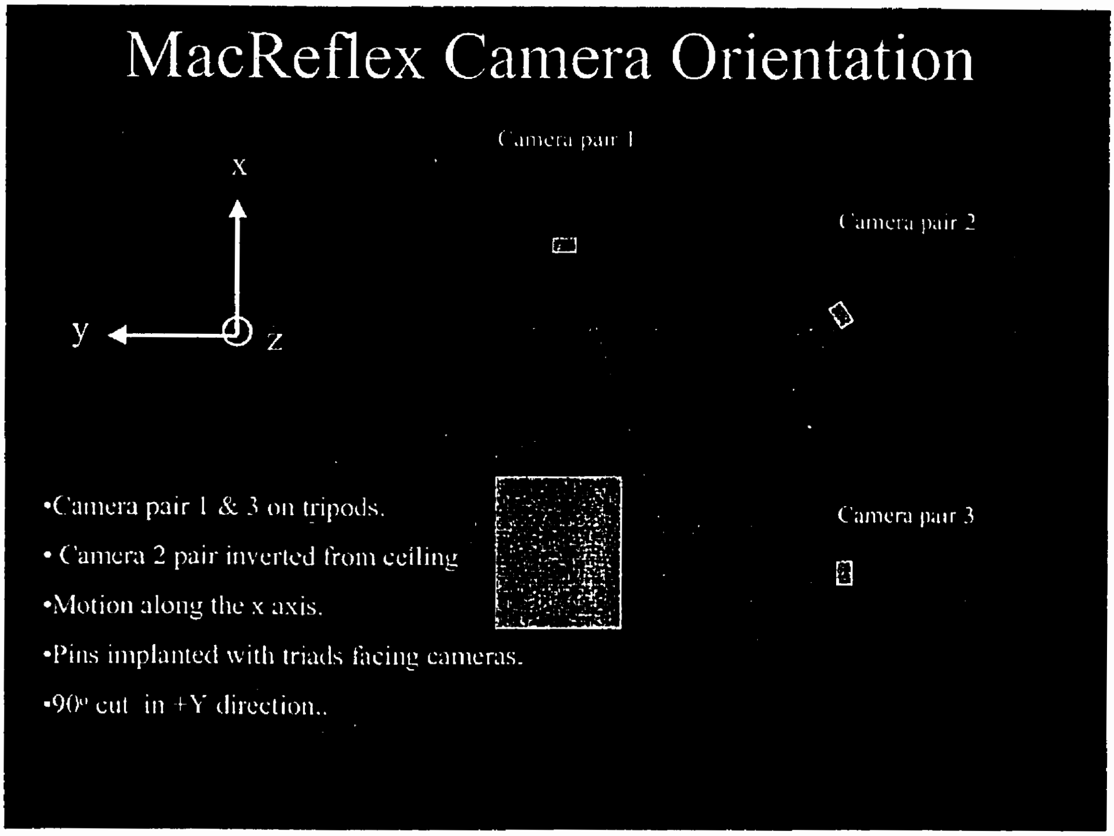
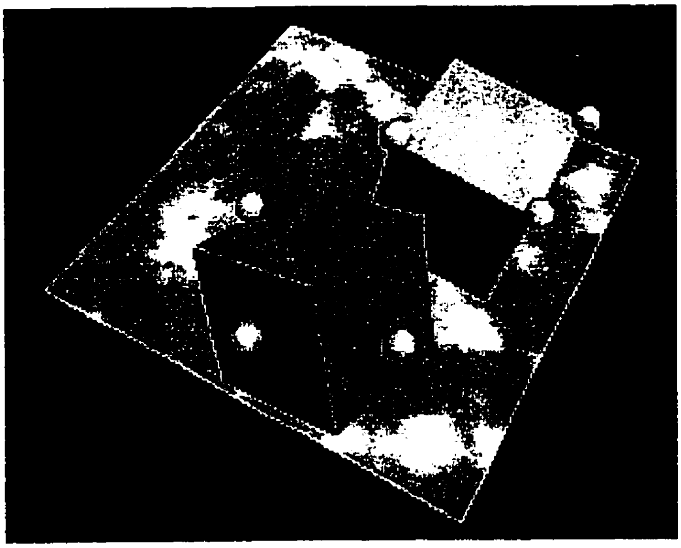


Figure B- 2: MacReflex camera orientation viewed from above

Appendix C

Segment Analysis



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The *Segment Analysis* software uses data from text-files (exported from the MacReflex motion analysis system) and performs the angular and translation calculations then writes the results in a new text-file. The process is described in Figure C-1.

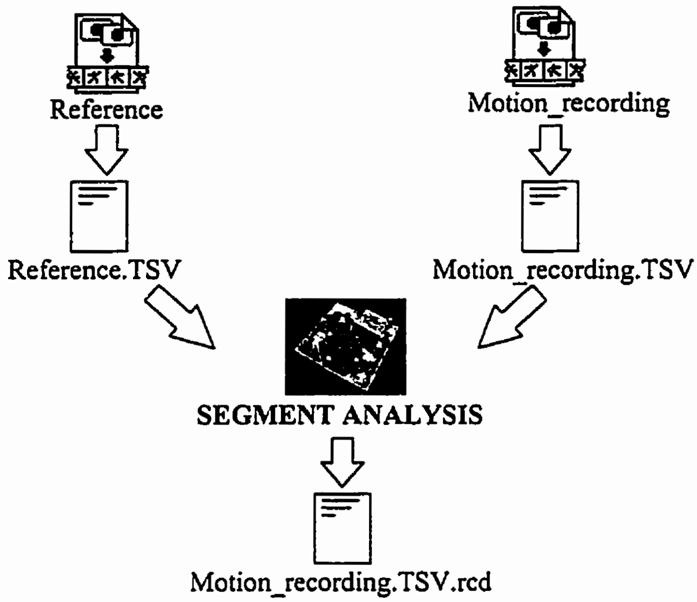

Figure C- 1: Data flow of the Segment Analysis software.

The analysis assumes the triads and associated markers (at least 3) have been attached to each segment of interest and their relative orientation and location remained constant during the recordings (in relation to the segment). The orientation of the target markers must remain fixed throughout the experiment to ensure accurate representations of 3D tibiofemoral motion.

Preparations

To use the *Segment Analysis* software, the least two segments must seen throughout the recording.

Prior to the actual motion recordings, a standing reference position must be recorded in a controlled posture whereby the orientation of the segments is in accordance with the laboratory coordinate system.

The segment of interest (tibia) is called the *measurement segment* of which 3-D motion relative to the *reference segment* (femur) is to be studied. With 3 markers attached to both the measurement and reference segments, a setup similar to the one shown in Figure C-2 can be found.

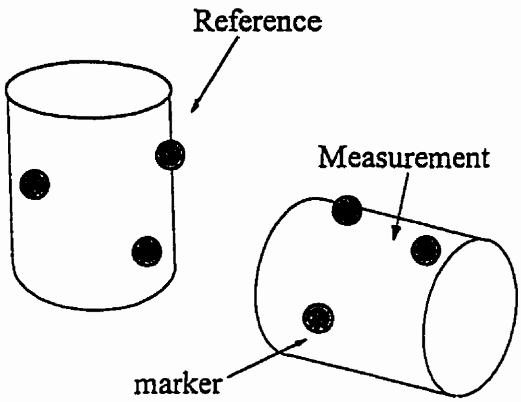

Figure C- 2: Markers attached to the reference and the measurement segment.

This position is recorded in order to obtain the necessary reference values as illustrated in figure 1. By using this recording as reference, the program will express the motions of the measurement segment in a co-ordinate system fixed in the reference segment. The axes of this system are oriented along the axes of the laboratory system when placed in the position recorded in the reference file (see Figure C-3).

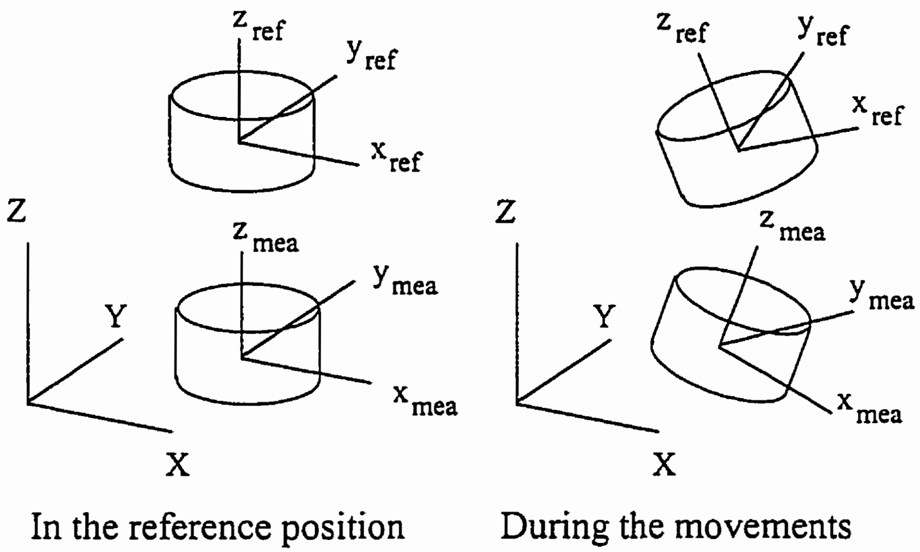

Figure C- 3: Orientation of the reference and laboratory system.

Analysis

Once you have performed appropriate MacReflex recordings, tracked your file and exported your data in the TSV-format, you are ready to begin segmental analysis. You must remember which marker numbers (of the tracked MaxReflex file) are attached to the measurement and the reference segments because this must be specified during the analysis.

To calculate **Relative translations,** the segments' x-, y- and z-coordinates for the tibia and femur including the fictive points for each segment must be specified. These may be obtained from stereo-photogrammetric-x-rays (RSA) of the segments. The dimensions must be expressed according to a right-oriented Cartesian coordinate system.

Description of the output files

The *Segment Analysis* program writes the output files in the same manner as the TSV-files, i.e. as text files with TAB as column delimiter. The content of the columns differ between the two kinds of files with the extensions *.and* or *.rcd.*

Angle analysis
The angle analysis output has the extension *.ang* and consists of four columns. The first one specifies the frame numbers (the same as the MacReflex file). The next three columns describes the three *alfa-*, *beta-* and *gamma*-angles which describes 3-D rotations of the measurement segment in relation to the reference segment. All units are in degrees.

The *alfa* angle is the rotation of the measurement segment that has occurred x-z-plane (of the reference segment). The alfa-rotation is positive about an axis parallel to the negative y-axis of the reference system. (positive values mean hyperextension of the knee).

The *beta* angle is the rotation of the measurement segment that has occurred y-z-plane (of the reference segment). The beta-rotation is positive about an axis parallel to the positive x-axis of the reference system. positive *beta* angles mean *adduction* of the lower leg in relation to the thigh (negative values mean *abduction*)

The *gamma* angle is the rotation of the measurement segment that has occurred x-y-plane (of the reference segment). The gamma-rotation is positive about an axis parallel to the positive z-axis of the reference system. a positive *gamma* angle means an internal rotation of the lower leg (negative values mean external rotation).

The angles are all set to zero when the measurement segment is oriented the same way as in Figure C-3 in relation to the reference segment.

Relative coordinates analysis
The relative co-ordinates analysis output has the extension *.rcd* and consists of 18 columns. The first one specifies the frame numbers (the same as the MacReflex file). The next three columns describe the x-y-z co-ordinates of the first measurement segment marker in relation to the reference system. The next three are for the next measurement marker and the next three for the last marker attached to the measurement segment. Columns 10 to 12 describe the x-y-z coordinates of the fictive point located somewhere on the measurement segment. Expressed another way, these columns represent the change of the tibial fictive point expressed in the femur-fixed system (using the reference position for zero-values). All coordinates are expressed in millimeters. Columns 13 to 15 are the Hx, Hy, and Hz coordinates as explained by Lafortune (1984; Lafortune et al., 1992). The remaining columns (16, 17 and 18) represents the results of the mediolateral shift, A/P drawer and compression-distraction based on McClay's (1990) clinical measures. The results are almost identical between columns Hx, Hy, and Hz and the last three columns but they may differ if a large ab/adduction should occur.

Calculations of Angles between segments

Sequence –y, x, z (Figure C-4)

The angles of the distal segments (in relation to the proximal one) can be calculated from the following equations:

$$\alpha = -\sin^{-1}\left[\frac{e_z^D \cdot e_x^P}{\cos\beta}\right]$$

$$\beta = -\sin^{-1}\left[e_z^D \cdot e_y^P\right]$$

$$\gamma = \sin^{-1}\left[\frac{e_x^D \cdot e_y^P}{\cos\beta}\right]$$

This approach is used by the Segment Analysis program and was originally developed for angle descriptions of the shoulder joint. See for example:

Karlsson D. and Lundberg A. (1994) In vivo measurement of the shoulder rhythm using external fixation markers. 3rd Int Symp on 3-D Anal. Of Human Movement. Hasselbacken Conference Centre, Stockholm. Proceedings: 69-72

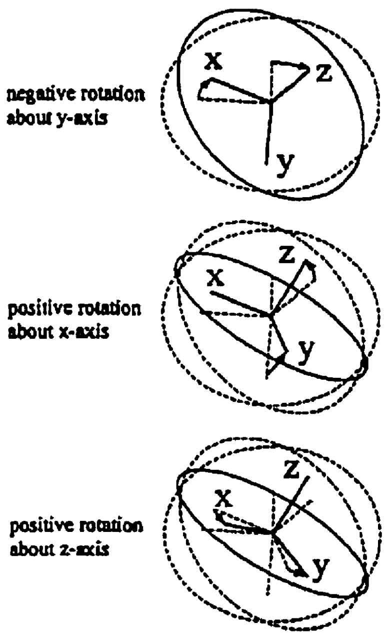

Figure C- 4: Angular descriptions as employed by Segment Analysis

A similar approach used in gait analysis

Sequence y, x, z (Figure C-5)

The angles of the distal segments (in relation to the proximal one) can be calculated from the following equations:

$$\alpha = \sin^{-1}\left[\frac{e_z^D \cdot e_x^P}{\cos\beta}\right]$$

$$\beta = -\sin^{-1}\left[e_z^D \cdot e_y^P\right]$$

$$\gamma = \sin^{-1}\left[\frac{e_x^D \cdot e_y^P}{\cos\beta}\right]$$

This approach was for example used in the following studies:

Kadaba, M.P., Ramakrishnan, H.K. and Wooten, M.E. (1990) Measurements of lower extremity kinematics during level walking. J. Orthop. Res. 8, 383-392.

Davis, R.B. Õunpuu, S. Tyburski, D. and Gage , J.P. (1991) A gait analysis data collection and reduction technique. Human Mvmt Sci 10. 575-587

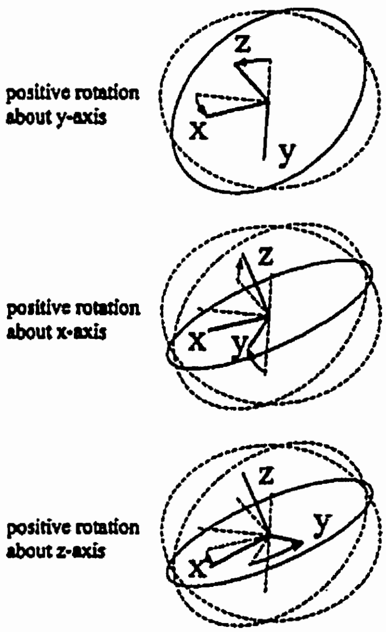

Figure C- 5: Angular descriptions as employed by Davis *et al.*, (1991)