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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

Objective. To describe three-dimensional tibial and femoral movements in vivo and examine the effect of a brace on knee kinematics during moderate to intense activity.

Design. Skeletal kinematics of anterior cruciate ligament deficient knees was measured with and without braces during moderate to intense activity.

Background. Invasive markers implanted into the tibia and femur are the most accurate means to directly measure skeletal motion and may provide a more sensitive measure of the differences between brace conditions.

Methods. Steinmann traction pins were implanted into the femur and tibia of four subjects having a partial or complete anterior cruciate ligament rupture. Non-braced and braced conditions were randomly assigned and subjects jumped for maximal horizontal distance to sufficiently stress the anterior cruciate ligament.

Results. Intra-subject peak vertical force and posterior shear force were generally consistent between conditions. Intra-subject kinematics was repeatable but linear displacements between brace conditions were small. Differences in angular and linear skeletal motion were observed across subjects. Bracing the anterior cruciate ligament deficient knee resulted in only minor kinematic changes in tibiofemoral joint motion.

Conclusion. In this study, no consistent reductions in anterior tibial translations were observed as a function of the knee brace tested.

Relevance

Investigations have reported that knee braces fail when high loads are encountered or when load is applied in an unpredictable manner. Questions remain regarding tibiofemoral joint motion, in particular linear displacements. The pin technique is a means for direct skeletal measurement and may provide a more sensitive measure of the differences between brace conditions.

Keywords: Knee joint; Tibiofemoral kinematics; Three-dimensional kinematics; Anterior cruciate ligament; ACL injury; Knee brace; Knee instability; Bone pin

1. Introduction

The criterion for determining whether anterior cruciate ligament (ACL) reconstructive surgery is required is based on patients' functional instability, from physical examinations and instrumented tests, e.g., KT1000 arthrometer (MEDmetric Corporation, San Diego, USA). Alternatively, functional knee braces are supposed 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 moderate–high physical activity. Braces are effective in reducing anterior translations when subjected to static or low anterior shear forces but fail in situations where high

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loads are encountered or when the load is applied in an unpredictable manner [1–6]. Since braces are designed for athletic activity, they should be evaluated under such conditions.

Knowledge about skeletal tibiofemoral joint motion is limited, in particular for secondary rotations and linear translations. Only recently it has been possible to directly measure the human tibiofemoral joint in vivo using markers attached to bone pins [7–12]. However, these studies have been restricted to semi-static activities, or walking and light running. There are no reports in the literature of application of this technique in examining the relationship between functional knee bracing and their effect on skeletal tibiofemoral joint motion during moderate to intense activity. Invasive markers implanted into the tibia and femur are the most accurate means to directly measure skeletal motion and this procedure may provide a more sensitive measure of the differences between bracing conditions. Therefore, this study utilised intra-cortical pin implantation and three-dimensional motion analysis to measure tibiofemoral kinematics in patients with ACL deficiency. The aim was to accurately measure the relative three-dimensional angular and linear movements between the tibia and femur and to investigate the effect of a knee brace on skeletal tibiofemoral kinematics during moderate to intense activity.

2. Methods

2.1. Subjects

Six male subjects with ACL deficient knees and having no prior surgical treatment 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. Deficient knees scored +2 on the Lachman's test and were evaluated with the KT 1000 arthrometer (MEDmetric Corporation, San Diego, USA) and compared against their contralateral leg. The ethics committee of the Karolinska Hospital approved the experimental procedure. Participants signed an informed consent form to participate in the study.

Of the six patient's, the results of four are presented. Two subjects were excluded, one because of iliotibial band impingement with the femoral pin the other the result of corrupted kinematic data. Personal data of the four patients are presented in Table 1.

2.2. Surgical procedure

Prior to surgery, the DonJoy Legend knee brace (Smith & Nephew DonJoy, Carlsbad, USA) was sized and fitted by the researcher as prescribed by the manufacturer. Both femoral and tibial insertion sites were identified so that no impingement occurred between the brace and Steinmann pins. Insertion sites were anterior and superior to the lateral femoral condyle and anteromedial to the tibial shaft. The brace was removed prior to the surgery.

Data collection follows an established protocol that has been fully detailed elsewhere [8–11,13,14]. In brief, the skin, subcutaneous tissue and periosteum were anaesthetised with standard anaesthetic. Steinmann bone pins (2.5 mm diameter) were inserted into the femur and tibia of each subject's deficient leg using a manual orthopaedic drill. To minimise impingement problems with the iliotibial band, the knee was flexed 45° prior to pin implantation [9]. The femoral pin was implanted superior to the lateral femoral condyle and directed obliquely in a posterior-medial direction. The tibial pin was inserted into the medial aspect of the tibial shaft and directed postero-laterally. Target clusters were then affixed to the pins. Each target marker was comprised of four non-collinear 7-mm reflective markers, one in the centre and three attached to orthogonal projecting rods. Since the anaesthetic was generally active for 2 h, this left ample time for the motion recordings. The pins remained inserted for the duration of the test.

Roentgen-stereophotogrammetric X-rays (RSA) were taken with the implanted pins to record the position of

Table 1
Subject data and knee scores including means and SD of the four subjects tested

	Subjects				
	A	B	C	D	Mean (SD)
Age (years)	29	19	19	22	22.3 (4.7)
Mass (kg)	75	79	72	90	79.0 (7.9)
Height (cm)	176	180	175	180	177.8 (2.6)
Lysholm knee score	72	75	75	74	74.0 (1.4)
Tegner activity score	4	9	6	6	6.3 (2.1)
Deficient leg	Left	Left	Left	Right	
Comparative KT 1000 data	9 mm	5 mm	4.5 mm	5.5 mm	
Brace application (trial)	Second	First	Second	First	

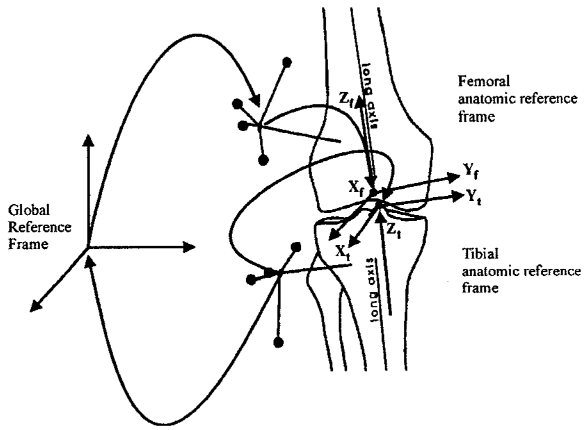

Fig. 1. Anatomical reference frame for the femur and tibia.

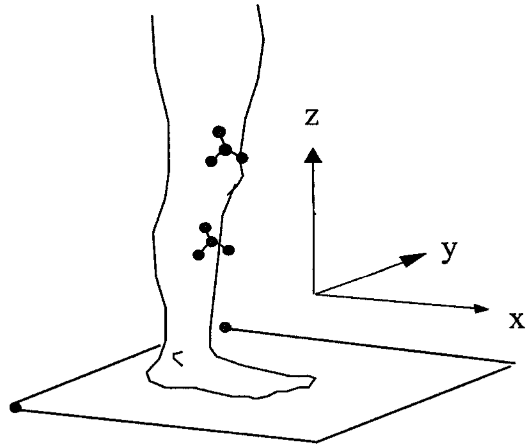

Fig. 2. Orientation of the lower leg and foot with reference to the force platform and the global coordinate system (adapted from *Segment Analysis* manual © Karlsson, 1997, p. 9).

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 origin of femoral and tibial anatomical reference frames were located at the deepest point of the intercondylar groove and on the most proximal point of the medial intercondylar eminence, respectively. In the right leg, the *Z*-axis was directed superiorly and parallel to longitudinal axes of the femur and tibia; the *X* was perpendicular to the *Z*-axis and progressed from posterior to anterior; and the *Y*-axis mutually orthogonal to both "*Z*" and "*X*" axes directed from lateral to medial (Fig. 1). For the right leg, a right-handed coordinate system was employed. The difference between the right and left leg was accounted for by manually negating the *Y* coordinates in the left leg and utilising the left-handed coordinate system to describe rotations [9].

2.3. Motion recordings

Six 60 Hz MacReflex infrared cameras (Qualisys, Sävedalen, Sweden) 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, or equivalent to three twin cameras sampling at 120 Hz. Each camera was equipped with 28° optical angle lens. The cameras captured all the markers about the knee from about footstrike through to knee extension. Prior to recording, a calibration frame with nine control points (volume $0.24\ \text{m} \times 0.16\ \text{m} \times 0.26\ \text{m} = 0.01\ \text{m}^3$) 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. Ground reaction forces (GRF) were simultaneously collected with a Kistler force plate (Kistler Instruments AG, Winterhur, Switzerland) sampling 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 orientated 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 (Fig. 2).

2.4. Experimental protocol and setup

Prior to surgery, patients completed the Lysholm Functional Knee Score [15] to assess their loss of knee function and the Tegner Activity Score [16] that ranks activities according to how troublesome they are to perform. Activity levels were later analysed in relation to the Lysholm Knee Score.

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 unsupported condition. For the braced trials, the knee brace (DonJoy Legend) was carefully reapplied by the researcher as prescribed by the manufacturer while ensuring the pins and target clusters were not touched or moved (Fig. 3). No impingement occurred between the brace and Steinmann pins.

After pin implantation, subjects were given several trials to perform the *One Legged Jump* (OLJ) to familiarise themselves with the pins and testing protocol. To

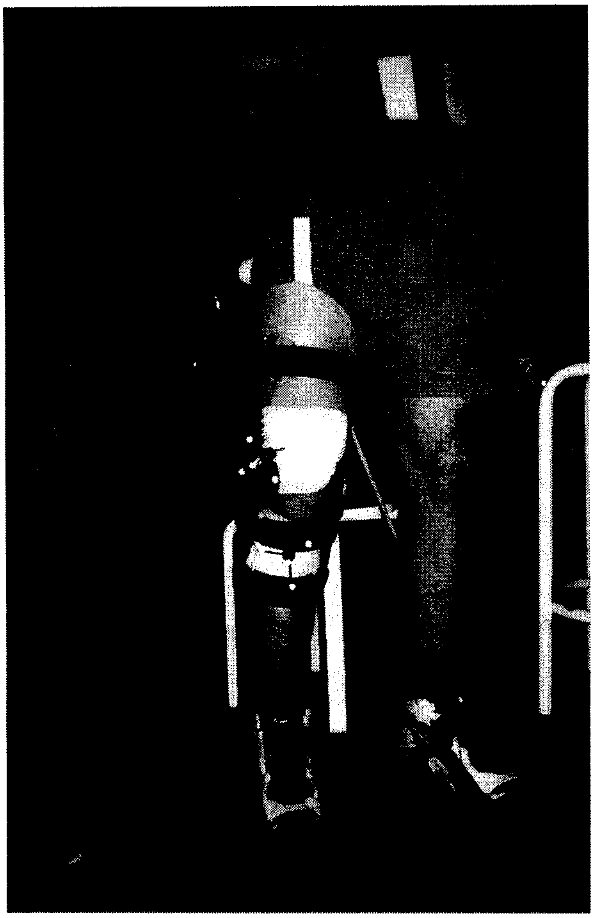

Fig. 3. Subject wearing the knee brace with target markers attached to the femoral and tibial pins.

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 onto their deficient limb. Their longest measurement was recorded and marked on the floor to determine the proper take-off distance to the force platform. After familiarisation with the procedure, a standing reference trial was recorded. 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 (Fig. 2). It was arbitrarily defined that the anatomical coordinate systems were aligned with the global coordinate system during standing. Five measurement trials and a second standing reference trial were recorded following the task. The orientation of the target clusters from the first trial was used as reference against the second to verify that the triad did not rotate on the pin during testing. When measurement recordings were completed for the first condition, standing and measurement trials were recorded for the second test condition.

The kinematic data were three-dimensionally reconstructed and angular rotations and tibial translations were calculated relative to the femoral anatomical reference frame using custom *Segmental Analysis* software, (Dan Karlsson, Frontec Research & Technology, Jönkoping, Sweden). The methods used to extract angular and linear data from coordinate transformation matrices are reported in greater detail elsewhere [8,9,13,14,17–19].

Angular descriptions were described using the conventions of Kadaba [20] and Davis [21] and computed using the rotation sequence about $-Y$, X, Z axes [22]. Tibiofemoral joint motion was described according to Grood and Suntay's joint coordinate system (JCS) [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 fourth-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 on both angular and linear data and by visual inspection. Initial contact with the force platform was noted to coordinate MacReflex (kinematic) data and GRF data. Kinematic and kinetic data for each subject and brace condition were time normalised to a specific interval. Briefly, the point when footstrike occurred was obtained from the force platform data and the corresponding frame number identified in the kinematic data. The jump sequence commenced 50 ms prior to footstrike through to knee extension and the associated posterior shear force began to plateau (Fig. 4). Average curves were derived using trials collected for each of four subjects during unbraced and braced testings. 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. Differences in the shape of the movement curves were reported rather than the absolute positions, i.e., ranges of motion (RoM) instead of absolute values.

Subjects ground reaction forces measured from the force platform were scrutinised and used as control between brace conditions. Vertical and posterior shear forces were scaled to body weight (including the brace when applicable) and time normalised following the criteria used to normalise the MacReflex (kinematic) data. All force data were interpolated with the coincident kinematic frame number so that each frame had a corresponding ground reaction force. If peak vertical force and peak posterior shear force imparted on the foot are similar between bracing conditions, then differences in translations may be attributed to the brace and not to differences in the landings onto the force platform. It was arbitrarily defined that peak vertical

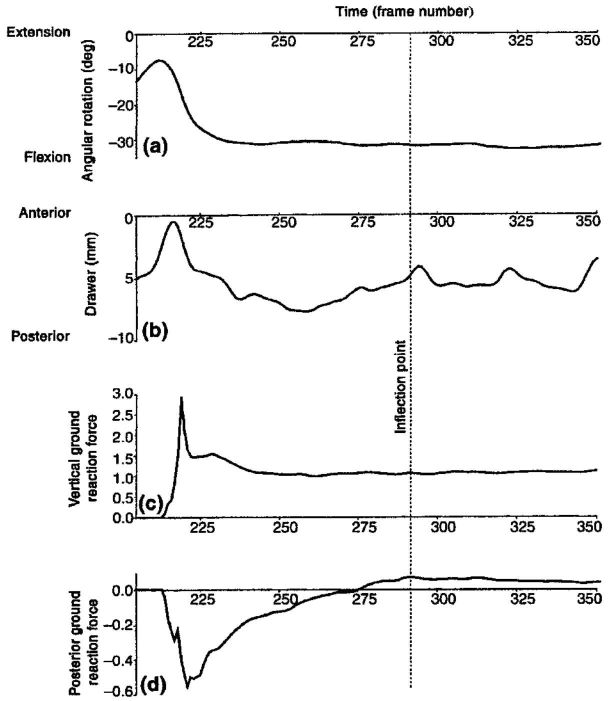

Fig. 4. A single trial that illustrates the One Legged Jump cycle. (a) Flexion/extension, (b) Drawer, (c) Vertical ground reaction force, (d) Posterior shear force. The inflection line identifies the plateau for posterior shear force, which is the endpoint used for time normalisation.

forces within 0.5 times bodyweight and peak posterior shear forces less than 0.3 were considered similar. Since the knee can experience forces up to eight times bodyweight, this criterion was considered to result in no mechanical or clinical significance.

3. Results

Of the six subjects, two were removed from the study. One subject generated enough force to bend the femoral pin approximately 10° during knee flexion, the result of an interaction with the soft tissue and musculature. For the remaining subjects, larger incisions were made about the femoral insertion site and deep flexion movements were avoided. The second exclusion was due to noise in the kinematic data that rendered linear interpolation impossible. No subjects experienced significant discomfort and all reported they could move their knees freely and their ability to jump was unaffected by the pins.

The Lysholm Functional Knee Score averaged across six subjects was 72.5 (SD, 2.6) ranging from 69 to 75. The mean Tegner Activity Score was 6.0 (1.9) and ranged from 4 to 9. All subjects had reported a difficulty in sport and the low Activity Score reflected this. None of the subjects had difficulty during daily activity as indicated by the moderate Lysholm Knee Score.

3.1. Ground reaction forces

Table 2 depicts mean peak vertical force (F_y) and mean peak posterior shear force (F_x) calculated from each subject's respective unbraced and braced trials. During unbraced testing, the data recording system failed to store ground reaction force data for subject 6. Mean peak vertical force ranged from 2.2 (0.27) to 3.4 (0.36) BW when unbraced and 2.4 (0.08) to 2.6 (0.59) BW during bracing. Mean peak posterior shear forces ranged from −0.6 (0.16) to −1.2 (0.17) BW and −0.6 (0.07) to −1.1 (0.11) BW when non-braced and braced, respectively.

3.2. Angular motion

Standard deviations of the differences between RSA and MacReflex values are reported to be less than 0.6° for rotations and less than 0.4 mm for translations when recorded in a volume of 0.01625 m^3 [24].

Average angular curves are illustrated in Fig. 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. Two subjects exhibited greater flexion (RoM) when the knee was braced (Table 3). Following peak flexion, three subjects stabilised the knee and remained in flexion overall whereas one subject returned to full extension (Fig. 5). Ab/adduction patterns were similar between bracing conditions but varied across subjects. Three subjects slightly adducted the lower limb (foot brought in towards the midline of the

Table 2
Mean peak vertical and peak posterior ground reaction forces normalised to body weight and mass of the brace across subjects and conditions

Subject	Trials	Peak vertical force (F_y)		Peak posterior shear force (F_x)	
		Unbraced	Braced	Unbraced	Braced
A	$n = 5$	2.9 (0.45)	2.6 (0.15)	−1.2 (0.17)	−1.1 (0.11)
B	$n = 3$	2.2 (0.27)	2.4 (0.08)	−0.6 (0.16)	−0.9 (0.09)
C	$n = 5$	3.4 (0.36)	2.6 (0.59)	−0.7 (0.07)	−0.6 (0.07)
D	$n = 5$	n/a	2.8 (0.30)	n/a	−1.1 (0.0)

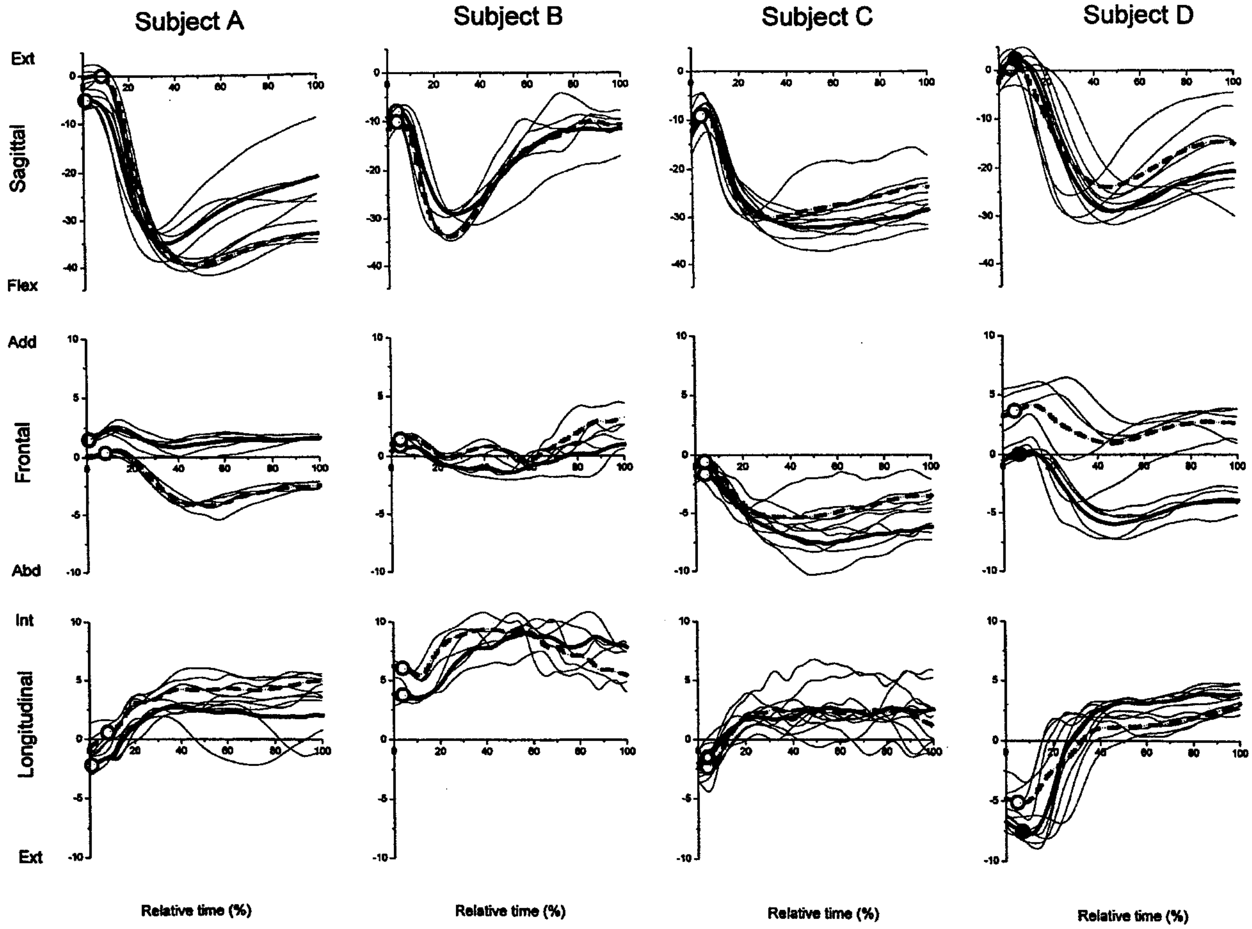

Fig. 5. Skeletal knee joint rotations for each subject. Means are displayed as a solid bold line for the unbraced kinematics, the bold dashed line represent braced kinematics. Footstrike identified as an open circle derived from force platform. Closed circle derived from kinematics.

Table 3
Means of angular ranges of motion[a]

Subject	Trials	Flexion		Abduction		Internal rotation	
		Unbraced	Braced	Unbraced	Braced	Unbraced	Braced
A	$n=5$	−29.9	−39.9	−1.5	−4.7	4.8	3.6
B	$n=3$	−21.1	−23.7	−1.4	−2.1	5.3	4.4
C	$n=5$	−24.2	−21.3	−5.9	−4.8	4.0	4.8
D	$n=5$	−31.5	−24.6	−6.2	−3.2	10.8	5.8

[a] Units in degrees: (i) A negative value indicates that flexion of the TFJ took place; (ii) A negative value indicates TFJ abduction and (iii) A negative value indicates external rotation of the TFJ.

body) until about peak F_y. Conversely, one subject abducted the tibia immediately at footstrike. Thereafter, all subjects abducted the tibia until about peak flexion. Abduction RoM between conditions and across subjects is listed in Table 3. When the knee was supported, abduction RoM was reduced in two subjects. The internal/external rotational patterns were fairly similar between conditions and across subjects with consistent RoM magnitudes except for subject D. Two demonstrated a small external knee rotation from footstrike to approximately peak F_y and all experienced a pronounced internal rotation throughout flexion. With the knee supported, three subjects had their internal rotation magnitudes reduced.

3.3. Linear displacements

Average joint translations for unbraced (solid bold line) and braced conditions (solid dashed line) are depicted in Fig. 6. Tibial displacements are described as movement of the tibial anatomical reference point relative to the femoral anatomical reference point. Anterior drawer is described along the floating axis. As seen in Fig. 6, anteroposterior 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 footstrike to approximately peak F_y (refer to Fig. 4). Thereafter, the tibial reference frame was drawn posteriorly during flexion. During bracing, anterior displacements remained unchanged for one subject; two subjects demonstrated small reductions while anterior displacements were larger for the remaining subject (Table 4).

The least amount of movement excursions was mediolateral shift. Average intra-subject patterns were similar in shape between bracing conditions although magnitudes varied considerably (Table 4). Entirely different movement patterns were observed across subjects. Two subjects demonstrated an initial lateral tibial shift from footstrike until about peak F_y. Thereafter, the tibia moved medially until peak flexion and remained constant thereabout. Two subjects exhibited small medial shift excursions from 20–40% after footstrike. Magnitudes were reduced for one subject between the unbraced and braced conditions.

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 Fig. 4, distraction–compression curves were similar in shape across bracing conditions. Additionally, distraction–compression patterns for three subjects exhibit a striking similarity with knee flexion–extension. Conversely, one subject exhibited contrasting patterns. As the knee extended prior to footstrike, little or no compression was evident for three subjects (Table 4). Dis-

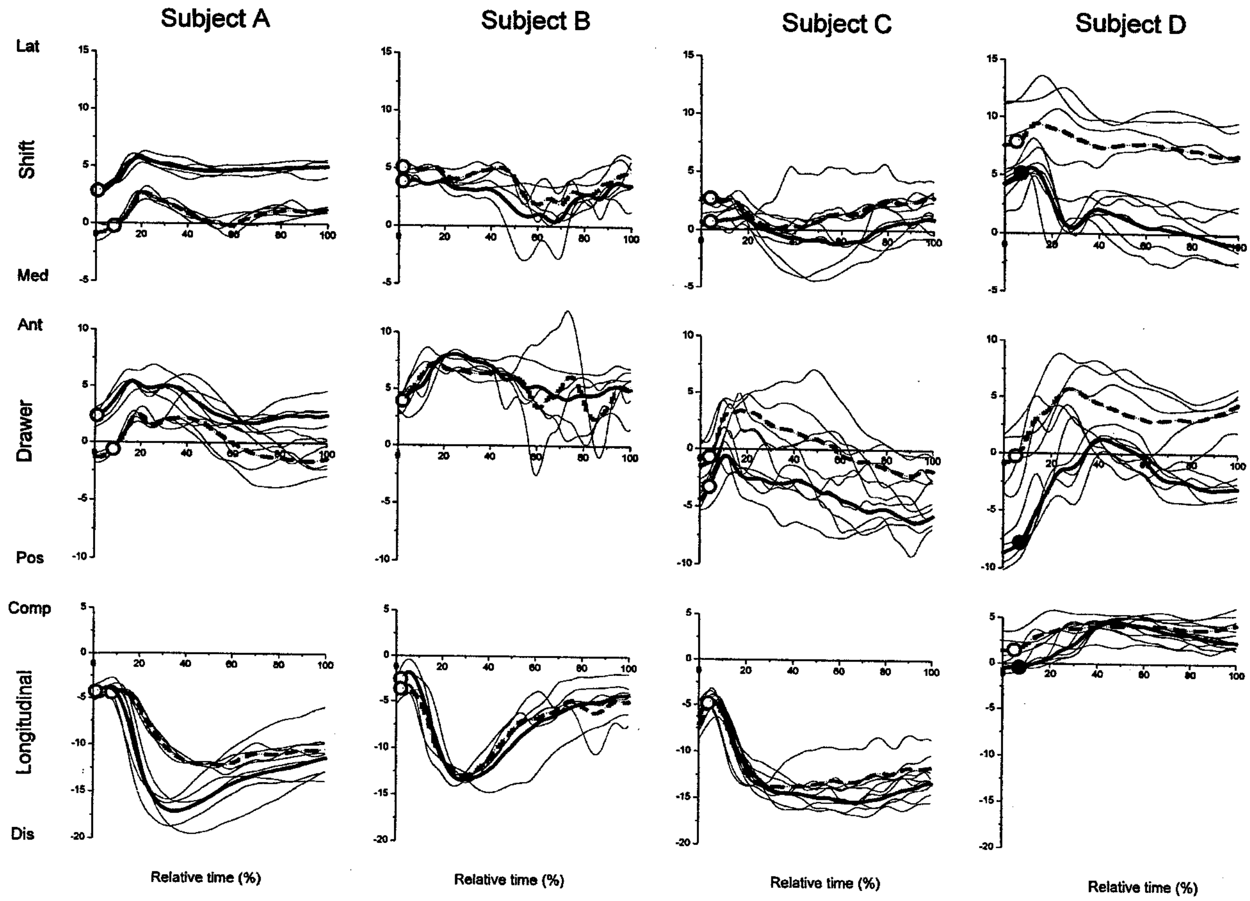

Fig. 6. Skeletal knee joint translations for each subject. Means are displayed as a solid bold line for the unbraced kinematics, the bold dashed line represent braced kinematics. Footstrike identified as an open circle derived from force platform. Closed circle derived from kinematics.

Table 4
Means of linear ranges of motion[a]

Subject	Trials	Medial shift		Anterior drawer		Distraction	
		Unbraced	Braced	Unbraced	Braced	Unbraced	Braced
A	$n = 5$	−1.2	−2.8	3.0	2.7	−12.9	−7.8
B	$n = 3$	−3.5	−3.1	3.5	2.4	−11.5	−9.2
C	$n = 5$	−2.2	−2.5	2.2	3.5	−10.8	−8.9
D	$n = 5$	−5.1	−2.1	8.8	5.7	5.0	2.6

[a] 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 and (iii) A negative value indicates that the joint was still compressed even though it was in its most distracted position.

traction occurred from footstrike until peak flexion followed by compression accompanying knee extension. Bracing reduced distraction magnitudes despite knee flexion magnitudes being larger for Subject A. The subject with the contrasting patterns experienced compressions during flexion with distractions during extension. A reduction of distraction–compression magnitudes was observed during bracing.

4. Discussion

Many investigations have reported knee braces to be ineffective during moderate to intense activity [1–6]. Additionally, little is known about effects of knee braces on three-dimensional osteokinematics, particularly skeletal tibiofemoral translations. Only recently it has been possible to directly measure human tibiofemoral joint motion in vivo using markers attached to bone pins [7–12]. Implanting intra-cortical pins into the tibia and femur is the most accurate means to measure three-dimensional tibiofemoral articulation and the pin technique may provide a more sensitive measure when examining the effect of knee bracing on skeletal tibiofemoral kinematics. Standard deviations of the differences between RSA and MacReflex values have been reported to be less than 0.6° for rotations and less than 0.4 mm for translations when recorded in a volume of 0.01625 m^3 [24]. Although this method is highly invasive and may cause undue discomfort, subjects from previous bone pin investigations reported they did not experience significant discomfort, they moved their knees freely and their walking and running styles remained unaffected.

For this investigation, intra-cortical pins were successfully implanted into patients with ACL deficiency to measure tibiofemoral joint motion, in particular anterior tibial subluxations during a dynamic activity. Additionally, this study has gone some way towards understanding the relationship between functional knee bracing and their effect on skeletal tibiofemoral joint motion during moderate to intense activity. However, due to the invasiveness of the protcol and since subject's jumped onto their deficient limb, jumps were within the patient's comfort limits.

Average peak vertical force at footstrike and peak posterior shear force was generally consistent between unsupported and braced conditions. This consistency indicates that landings onto the force platform were similar and changes in skeletal kinematics cannot be attributed to differences in landings but rather to the brace itself. However magnitudes varied across subjects owing to the fact that subjects jumped within their own comfort limits.

Overall, tibifemoral rotations and translations show a general trend across subjects, i.e., the shape and amplitudes of the skeletal marker based curves were fairly similar. Intra-subject differences between unbraced and braced patterns were small, i.e., knee kinematics were very repeatable. Although this study included a small number of subjects, the results show quantitatively for the first time that bracing the ACL deficient knee resulted in only minor kinematic changes in tibiofemoral joint motion. The inter-subject differences were typically much larger with differences mainly consisting of amplitudes and positional changes at touchdown.

Ab/adduction RoM is limited to approximately 5° due to ligamentous and geometry of the knee [25]. The range of motion may even be smaller during high dynamic activity since the knee is loaded and stabilised by muscular forces [14]. The large magnitude of adduction found for subjects A, C and D may not reflect "true" ab/adduction patterns. Secondary rotations are highly susceptible to cross-talk from flexion–extension [14]. These stem form alignment problems of the anatomical coordinate systems the result of which movements may exceed and mask the actual motion. Ab/adduction patterns for subjects A, B and D were very similar to the flexion–extension motion, giving rise to speculations that the relatively large abductions might be caused by cross-talk.

With respect to linear displacements, the tibia exhibited a rapid anterior drawer relative to the femur from footstrike to approximately when the peak vertical force was reached. Thereafter, the tibia was drawn posteriorly during flexion. The negligible reduction in anterior drawer between brace conditions indicates the brace did not reduce translations during dynamic activity. An offset between the unbraced and braced trials was evident in the recorded data. This may be due to the brace but is more likely the result of the different standing reference trials used for each condition. This created small deviations in alignment of the tibial and femoral anatomical coordinate systems [14]. It must be realized that some of the angular and linear differences evident across subjects be attributed to inconsistencies in defining the anatomical coordinate systems and cross-talk.

Comparisons with other intra-cortical research is problematic because of the different recording methods and the different definitions of the femoral and tibial anatomical reference points [8–11,14]. Lafortune [8] and McClay [9] employed anatomical reference based on a Roentgen-stereophotogrammetric analysis whereas Reinschmidt [10,11,14] utilised the neutral standing trial. Murphy [7] used the concept of instantaneous helical axes. Since rotations about and translations along a unique spatial axis have no anatomical references, no translations can be described. Ishii [12] measured tibiofemoral joint motion with an instrumented linkage system with intra-cortical Kirchner fine wires between full extension and 60° flexion while the subject was seated. Also, it would be expected that patterns and magnitudes may vary dependent on the activity involved, particularly walking compared with running.

Nevertheless, tibiofemoral translations were in general agreement with Reinschmidt [14]. Reinschmidt related flexion with a lateral shift, a posterior drawer and a tibial distraction while extension demonstrated contrary patterns. Conversely, Lafortune [8] and McClay [9] associated flexion with a medial shift, a posterior drawer and a tibial distraction with the opposite being true for extension. McClay [9] found similar anterior/posterior patterns during running although magnitudes were larger. 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. Furthermore, Lafortune [8] and McClay [9] described linear displacements as absolute values relative to the positions of the tibia and femur at heel strike. Reinschmidt [14] reported translations as changes in movement between the origin of the femoral and tibial anatomical coordinate systems already some distance apart.

In this study, 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 [26]. However, during all motion recordings, the markers remained within the calibrated volume. Additionally, cardan angles and joint translations calculated using the JCS are highly susceptible to alignment errors and uncertainties in defining the anatomical coordinate system [14]. Small deviations in alignment of the anatomical frames of reference across subjects make inter-subject comparisons difficult. Reinschmidt [14] indicated that translations might be dependent on the rotations. Blankevoort [27] suggested meaningful distances to 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 JCS.

5. Conclusion

The techniques used in this investigation have provided a valuable insight into the three-dimensional kinematic behaviour of the ACL deficient knee joint during moderate to intense activity. Although this study included a small number of subjects, the results show quantitatively for the first time that bracing the ACL deficient knee resulted in only minor kinematic changes in tibiofemoral joint motion. No consistent reductions in anterior tibial translations were observed as a function of the knee brace tested.

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