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The Normal and ACL Deficient Knee: An *In-vivo* Three Dimensional Kinematic and Electromyographic Analysis

av

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AKADEMISK AVHANDLING

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Cover: Subject with brace and target markers affixed to bone pins. X-ray of the knee depicting the femoral and tibial anatomical reference points, frontal view. Photos and illustrations of the knee created by Dan K Ramsey. Illustrations reprinted with permission from Elsevier Science.

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

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ABSTRACT

The overall aim of this thesis was to describe the *in-vivo* 3D kinematics of the normal and ACL deficient knee. Baseline data from normal controls were compared against the pathological knee to assess whether generic functional knee braces reduced abnormal displacements. Moreover, 3D bony contact movements were derived from MRI data.

Six patients with ACL rupture and 8 normal controls participated in the invasive pin experiments. During dynamic activity, 3D motion analysis was used to record knee kinematics using markers affixed to 3.2 mm Hoffman bone pins inserted into the tibia and femur. Simultaneous neuromuscular activity was measured for the patient group. Ground reaction force (GRF) data were control for to ensure consistent jumps. For the MRI study, recordings were taken of the right knee of sixteen normal subjects.

For the pin studies, all subjects moved their knees freely. None experienced significant discomfort. For the patient group, intra-subject GRF data was generally consistent between bracing conditions. Therefore, landings were considered similar. Intra-subject kinematics was repeatable but no consistent reductions in anterior translations were evident. Anterior displacements when unsupported were on average 4.4 mm (range 2.2 to 8.8 mm) and 3.6 mm (range 2.4 to 5.7 mm) with the knee braced. Neuromuscular changes were observed as a function of bracing. Semitendinosus activity significantly decreased 17% prior to footstrike (P<0.05). Bicep femoris significantly decreased 44% whereas rectus femoris activity significantly increased 21% following footstrike (P<0.05). These findings suggest joint stability may result from proprioceptive feedback rather than the mechanical stabilising effect of the brace. Therefore, knee bracing combined with proprioceptive and muscular coordination training to increase joint stability is important. When normal controls were considered in combination with our brace study, peak vertical force values did not exceed 2 times bodyweight whereas the ACL deficient group experienced up to 3 times bodyweight. Displacement magnitudes observed for the non-braced and braced ACL deficient subjects were similar to normal controls. Mean anterior tibial displacements for normal controls were 3.7 mm (range 1.2 to 7.3 mm) and 6.1 mm (range 1.4 to 10.3 mm) for the jump and cut respectively. The level of activity may not have been adequate to provoke the large displacements in order to identify whether knee braces reduced pathological translations to within normal. For the MRI study, movement of the contact points between the medial and lateral condyles was posterior 12.5 mm vs. 13.8 mm, laterally 0.6 mm vs. 1.4 mm and inferiorly 6.2 mm vs. 9.7 mm when flexing to 30°. From 30° to 60°, the contact points moved anteriorly 1.0 mm vs. 6.4 mm, continued laterally 2.5 mm vs. 2.4 mm, and progressed superiorly 2.4 mm vs. 4.7 mm. Further research is required to verify these findings during weight-bearing conditions.

Problems were associated with the femoral pin. During mechanical testing, loads of 150 N and 100 N applied at 15 mm and 20 mm respectively produced deflections larger than 0.4mm. A resonance frequency of 90 Hz was observed. By improving insertion and pin design, this technique may become more reliable.
Keywords: Knee, Tibia, Femur, Tibiofemoral contact points, Tibiofemoral kinematics, Anterior cruciate ligament, ACL injury, Knee brace, Knee instability, Bone pin, Magnetic resonance imaging, MRI, Neuromuscular adaptation, Electromyography.

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List of Papers

The thesis is based on the following original papers. They will be referred to in the text by their respective Roman numerals.

I. **Assessment of functional knee bracing: An in-vivo three-dimensional kinematic analysis of the anterior cruciate deficient knee**.
Ramsey DK, Lamontagne M, Wretenberg PF, Valentin A, Engström B, Németh G.
Clinical Biomechanics 2001; 16: 1-70

II. **Electromyographic and biomechanic analysis of anterior cruciate ligament deficiency and functional knee bracing.**
Ramsey DK, Wretenberg PF, Lamontagne M, Németh G.
Clinical Biomechanics 2003; 18: 28-34

III. **A three dimensional kinematic profile of normal tibiofemoral joint motion during strenuous activity**.
Ramsey DK, Wretenberg PF, Németh G.
In manuscript, 2002

IV. **Methodological concerns using intra-cortical pins to measure tibiofemoral kinematics**.
Ramsey DK, Benoit D, Wretenberg PF, Lamontagne M, Németh G.
Submitted for publication, 2002

V. **Tibiofemoral contact points relative to flexion angle measured with MRI**.
Wretenberg PF, Ramsey DK, Németh G.
Clinical Biomechanics 2002; 17: 477-485

CONTENTS

ABBREVIATIONS

ACL	Anterior cruciate ligament
Anterior drawer	Anterior tibial displacement relative to the femur.
EMG	Electromyography
Fy	Vertical ground reaction force.
GRF	Ground reaction force
IEMG	Integrated EMG
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.
MRI	Magnetic resonance imaging
OLJ	One legged jump
RoM	Range of motion
RSA	Roentgen-stereo-photogrammetry
3D	Three-dimensional

GENERAL BACKGROUND

Quantitative kinematic analysis is important for gaining a thorough understanding of normal and pathological knee joint function during human locomotion [108,110]. Developing normal joint profiles, accurate biomechanical models and identifying abnormalities makes it possible to further improve diagnosis and treatment. The design of rehabilitation programs and orthoses may also be enhanced.

Currently, no single methodology exists that produces optimal measuring techniques to record tibiofemoral joint motion. Although, an abundant amount of data has been collected employing different experimental designs, much of the results are contradictory and comparisons across investigations are nearly impossible [89,107,130].

For a complete kinematic analysis, three dimensional motion analyses is required whereby all six degrees of freedom (three rotations and three translations) can be discerned [17,18]. Early kinematic analysis was restricted to two-dimensional analysis due to the difficulty in obtaining motion relevant to six degrees of freedom [77]. With advances in computerized motion analysis, an accurate technique for documenting and developing normal 3D tibiofemoral joint profiles is now available [50].

Yet knowledge about skeletal tibiofemoral kinematics is limited since measurements have usually been accomplished by attaching reflective markers to the surrounding soft tissue of the calf and thigh [58]. Surface markers may not accurately represent the underlying bone motion during dynamic activity [108,110]. The relative movements between skin markers and the underlying bone may introduce large errors [96]. This is a particular concern during high dynamic activity. One way to avoid the problem is to directly measure skeletal motion *in vivo* [108,110]. However, the applicability of such methods is limited, mainly due to the invasiveness of such procedures.

KINEMATICS OF THE KNEE JOINT

Kinematics about the knee consist of three rotations; internal-external, abduction-adduction and flexion-extension and three translations; anterior-posterior drawer, medial-lateral shift, distraction-compression [77,88,89,125]. This terminology provides a clinical interpretation of the motion with translations referring to displacements with respect to the femur. Note, compression-distraction does not refer to physical compression of the contact areas between the articulating surfaces of the tibia and femur [77]. Rather it is analogous to the entire tibiofemoral joint being shortened or stretched along the longitudinal axis. As the knee flexes, the femoral anatomical coordinate system is drawn superiorly and away from the tibial anatomical coordinate system. The opposite holds true for extension.

Tibiofemoral contact areas and pressure distributions have been measured directly in cadaveric specimens employing various invasive techniques. These include measuring the radiolucent contact area from a Roentgenogram [72], using pressure sensitive film (FUJI Film Pressensor film) [10,40,49,52,56,113], or microindentation transducers [57,70]. Interpreting the results from these *in vitro* studies are limited as they are obtained under non-physiological loading conditions.

Pressure sensitive film has also been successfully applied *in-vivo* to measure cat patellofemoral joint contact stresses and areas [112]. However, insertion of transducers or Pressensor film into an articular joint disturbs and changes the contact mechanics of the joint so that the true contact pressures can never be measured experimentally [144]. Using conventional 2 dimensional x-rays, Nisell *et al.* [100] recorded average sagittal contact points for both the medial and lateral knee compartment during flexion.

From sagittal MRI images of unloaded cadaveric knees, Pinskerov *et al.* [103] and Iwaki *et al.* [59] described the femoral condyles as comprising two posterior circular arcs: a small anterior extension facet and a large posterior flexion facet. These arcs do not share similar radii in what the authors termed "kink angle".

Along the surface of the medial tibial plateau, the femur lies in a cup-shaped "recess" bounded anteriorly by a slope that is upward and forwards and posteriorly by a relatively flat flexion facet. Only the posterior tibial flexion facet contacts the femur directly, at about 20° to 120°. Laterally, the tibial condyle is convex but the central articular surface is relatively flat. Only the flat surface articulates with the femoral facets.

When the knee flexes from terminal extension, the femur initially rocks back onto the posterior flexion facet from the anterior flexion facet. Thereafter, knee flexion is a combination of the femoral condyles rolling over the tibial plateau and the posterior gliding of the condyles along the plateau. Translational assumes an increasing proportion of knee motion because of the shape of the femoral condyles [88]. Ab/adductor motion is limited to approximately 5° due to the restrictions imposed by the knee's geometry and the collateral ligaments [39].

METHODS IN RECORDING TIBIOFEMORAL JOINT MOTION

Early tibiofemoral kinematics has been measured using utilizing human cadaver specimens although the compliance of living tissue and the effects of active muscular contractions are ignored. The magnitude of active muscular activation and the sequence of muscle recruitment introduce large forces experienced by the ACL [3,12,14,38,111,139]

Electrogoniometers have been used in order to describe motion of the tibia with respect to a fixed femur during level walking [88,133,135]. However, these investigations have been subject to criticisms due to movement restrictions or motion artefacts imposed by the devices [17,46]. Later, Vergis *et al.* [134] reported good reproducibility using an electrogoniometric technique during measurements of anterior tibial translation relative to the femur.

More direct methods have employed arthrometers that register an applied anterior force while measuring associated tibial translations relative to the femur. Testing is usually restricted to static positions with anteriorly applied forces below 200 N (below the inherent stiffness of the soft tissue) leading to criticisms that forces must exceed 200 N with 400 N more representative forces acting on the knee during activity [23,125].

Conventional high speed film camera's (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 [76,77,78,89,96,108,109,110]. However, cinematography has its disadvantages as the process of manual digitisation further increases the chance of errors for determining the joint or marker centre [50,137]. For this reason, automated motion analysis systems have replaced film, the most common being passive systems [50,102]. 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 $2m^3$ volume [71] while other studies found errors of 1-3 mm using similar volumes [65,73]. Lundberg et al. reported errors than 0.6° for rotations and less than 0.4 mm for a volume of 0.01625 m^3 [86].

New non-invasive technologies have advanced the means by which skeletal tibiofemoral kinematics has been measured. These include roentgen-stereo-analysis [19,63,64,68,69,85], biplanar image-matching [7], video fluoroscopy [8,9,25,26,27,123,124,126] and cine PC-MRI techniques [117,118,119]. The main limitation of RSA has been the difficulty in examining the knee joint during continuous movement and weightbearing [19]. Moreover are the limited sampling rates. The challenge of video fluoroscopy is to identify the exact location of bony landmarks for every time frame in the x-ray views. Distortion may result when an image intensifier and recorder system are used in conjunction with the fluoroscopic system [136]. Additionally, the equipment used for data collection and the associated measurement volume is confining and does not allow for unconstrained movement, like performing daily activities or sport manoeuvres. Moreover, the low sampling rates are not suitable for dynamic activities and validation of these methods is difficult. Faster sampling times are associated with increased radiation exposure.

Magnetic resonance imaging (MRI) has proven to be an accurate non-invasive assessment tool and it has several unique advantages. It can be used for recordings of 3D co-ordinates and provide precise visualisation of muscles and tendons, bone, and volume reconstruction when analysing articular joints [33,98,99]. Various MR techniques have been employed such as passive techniques and incremental positioning of the segment during muscle relaxation [92,98,99]. Additionally, MRI has been used to measure passive muscular moment arms *in-vivo* [142]. The drawback is that standard MRI machines have relatively narrow imaging chambers. Consequently, the knee can only be imaged in non-weightbearing extension for living subjects. Open access MRI has made it easier to examine the knee in non-weightbearing flexion or during simulated weightbearing. In studying knee kinematics, MRI has offered new opportunities and applications in the diagnosis of knee disorders [92,98,99]. To date, investigators have begun to report on use MRI as an anatomic tool to study the articular surfaces of the femur and tibia and the relative positions of the condyles at various degrees of flexion [51,59,94,103]. However analysis was restricted to sagittal plane movement.

Alternatively, the method of implanting intracortical pins into the femur and tibia affixed with target clusters may provide the more sensitive measure of tibiofemoral joint motion during strenuous activity [58,77,81,89,93,106,107,109,110]. The method allows for very high spatial resolution ($< 0.6°$ and < 0.4 mm for a 0.01625 m^3 volume) and temporal resolution is limited by the motion capture system [86]. However, tibiofemoral kinematics measured with pins has 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 remains the most accurate and reproducible means to directly measure skeletal motion [21,96] and this procedure may provide a more sensitive measure of the differences between bracing conditions. However, application of this method is limited, mainly due to the invasiveness of such procedures, methodological considerations and problems with the femoral pin [107]. Additionally, knee joint contact points cannot be measured with this technique [58,77,109].

Markers and artifacts

Reflective markers placed on the body are in theory supposed to represent the position of anatomical landmarks for the segment in question [96]. However, surface markers may not represent the true anatomical locations resulting in *relative* and *absolute errors* [96]. Relative errors are movements between markers with respect to each other and are caused by skin movement relative to bone [58]. An absolute error is movement of a marker with respect to a specific body landmark [96]. 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 [58,96].

Skin movement artefact presents the most critical source of measurement error [20]. The relative motion between skin markers and the underlying bone may introduce errors of 10 - 40 mm, particularly at the thigh [67,83,96,109]. Therefore, ab/adduction and internal/external rotations about the knee cannot be accurately represented using surface mounted markers [20]. This is of particular concern since movement artefacts all but restrict kinematic analysis to two-dimensional sagittal analysis. Various methods have been employed to correct for movement artefacts, e.g. the application of clusters of markers or the solidification of marker groups [5,28,83]. The results from these techniques are encouraging with improvements of up to 33% accuracy [5]. However these methods have yet to be validated and therefore have limited applicability.

A way to avoid the problem of surface markers is to use invasive markers to directly measure skeletal motion [87]. This provides the most accurate means for determining bone movements [21,96]. Differences of up to 50% for similar knee angles when comparing tibiofemoral joint kinematics using external and bone fixed markers have been reported [96].

INTRACORTICAL PIN TECHNIQUE

A pioneer in the use of intracortical pins to study human motion *in-vivo,* Levens *et al*. [81] 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 [81].

Lafortune *et al.* [77] 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 Steinmann pins (2.5 mm diameter) affixed with target clusters into the adductor tubercle, into the lateral tibial condyle and into the midpatella [76,77,78]. Each triad contained four noncollinear spheres (Figure 1), one in the centre and three attached to orthogonal projecting rods [77].

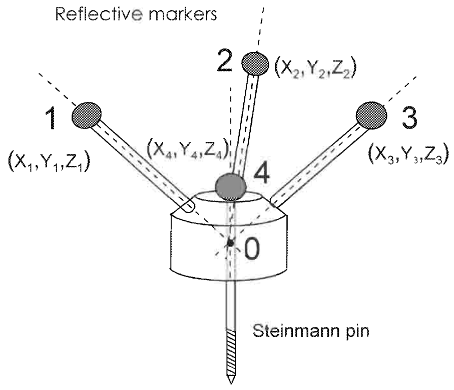


Figure 1: Schematic of target cluster.

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 [77]. Since these anatomical landmarks can be identified with great precision, an accurate description of skeletal movement is possible [108]. Orientation of the target markers remained fixed throughout the experiment.

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) [1]. A series of transformation matrices [79] 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 joint co-ordinate system of Grood and Suntay [47].

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 [76,77,78].

In an unpublished doctoral thesis, McClay [89] employed a similar protocol to examine tibiofemoral and patellofemoral kinematics of two non-injured runners and two patellofemoral pain sufferers. The femoral pin was inserted laterally with the knee flexed 45°. 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.

Later Reinschmidt et al., [109] compared skin marker and skeletal marker motion during the stance phase of walking and running[110]. Hoffman bone pins (2.5 mm in diameter) affixed with target markers 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 normalized 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 [108,110]. In contrast to neutral standing, RSA enables anatomical meaningful origins to be defined, which allow for joint translation determinations. Employing transformation matrices [121], 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 [47].

Reinschmidt [108] 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 [84,86,115,116,132], 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 [108].

To define both tibial and femoral anatomical reference frames, Lafortune *et al.* [77] McClay [89] and Reinschmidt *et al.* [108,110] 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.

The deepest point of the intercondylar groove was chosen as the origin for the femoral co-ordinate system (Figure 2). 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 [76,77,78,89,108,109,110].

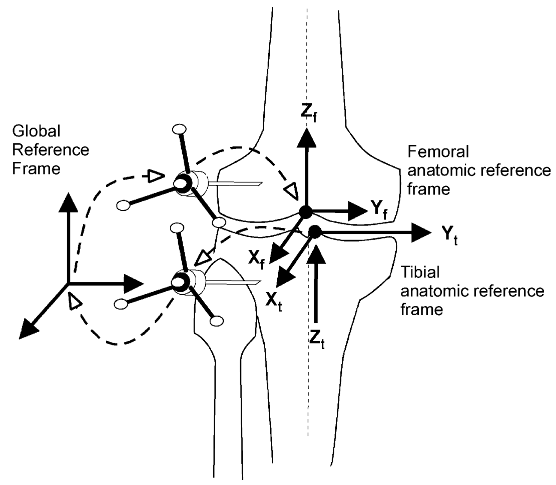


Figure 2: Anatomical reference frame for the femur and tibia.

Analytical methods to quantify joint motion

The joint co-ordinate system is the most common method and is based on local anatomic landmarks [24,47]. The joint co-ordinate system uses Cardanic (or Euler) angles with respective translations to describe joint motion about axes defined in the anatomical segments [108].

To define the anatomical co-ordinate system, methods include neutral standing [6,91,97,108,110], roentgen-stereo-photogrammetric analysis [76,77,78,89,108,109], and relationships between bone embedded reference frames and external markers placed on anatomical landmarks [22]. 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 tibial anatomical co-ordinate system into the femoral anatomical co-ordinate system [79,121,122].

For gait analysis in clinical settings, the most commonly used co-ordinate system is the "joint co-ordinate" system [47]. 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 [108]. 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 bilateral femoral axis, ab/adduction and anterior-posterior drawer around the floating axis and internal/external knee rotation around fixed tibia longitudinal axis.

Negligible effects using bone pins

No subjects have been reported to experience pain and/or significant discomfort during the experiments; all reported being able to move their knee freely despite pin implantation [77,78,89,109,110]. Although some problems were associated with the femoral insertion sites, most subjects engaged in normal activities two weeks following the experiment. In comparing skin marker kinematics for walking and running with and without bone pins, the similarity in the shape and amplitude of the curves suggests the bone pins did not affect walking and running styles [108].

Source of error: Anatomical co-ordinate system versus cross talk

When measuring three-dimensional motion *in vivo*, the choice of anatomical co-ordinate systems is of great importance [108]. 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 [108]. Ramakrishnan [104] 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 [108]. 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 [108].

Cross talk is primarily a concern for joints that rotates principally about one axis, such as flexion/extension of the knee [108]. 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 [108]. 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 [108,109]. 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*.

FUNCTIONAL BRACING TO STABILISE ACL DEFICIENT KNEES

Whereby the primary role of the ACL is to resist anterior-posterior translation, functional knee braces are supposed to provide stability to the ACL deficient knee by reducing pathological translations and rotations [130]. However, when engaged in athletic activities, subjects have reported continued episodes of the knee instability with the limb supported. Despite the amount of research examining the effectiveness of functional knee braces, much of it is contradictory and inconclusive.

Unlike prophylactic or rehabilitative braces, functional braces provide stability to an unstable knee [130]. 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 [130]. 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 [130]. Key to the braces rigidity is the straps or shell. The tighter and more rigidly the brace is applied, the better the match for knee motion [130].

Early studies reported few symptoms of instability with improved athletic performance although evaluations were anecdotal and subjective in nature [131]. Later braces have been reported to be effective in reducing anterior translations when subjected to static or low anterior shear forces. But they fail in situations where high loads are encountered or when the load is applied in an unpredictable manner [17,23,30,34,130,131]. which are thought to accelerate the degenerative joint disease seen in anterior cruciate deficient knees [17]. 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 stabilizing effect of a brace [131].

To date, little research has examined the effects of knee braces on three-dimensional osteokinematics and arthrokinetics during strenuous physical activity. Manual knee evaluations including the anterior drawer test, the pivot shift test, and Lachman's test, all have measured tibial displacements during simulated static loading conditions but they do not reflect true physiologic loading [17]. Since braces are designed for athletic activity they should be evaluated in such conditions. For example, to challenge the knee with muscle contraction and bodyweight to increase ACL strain in loading situations where patients depend on the brace [13,15]; to produce abnormal anterior tibial translations and to include stiffness of the soft tissue to examine the braces ability to protect articular structures [15]. With the proliferation of new functional braces claiming to stabilize ACL ligament deficient knees, clinical and laboratory research is necessary to substantiate their effectiveness [17,18,30].

Six degree of freedom electrogoniometers have been used in order to describe motion of the tibia with respect to a fixed femur. Marans *et al.,* (1989) reported significant differences in the shape and magnitude of the anterior/posterior translation curve for a group of normal and ACL injured subjects. Two distinct patterns were observed: ACL deficient subjects exhibited increased amplitudes during swing and consistent decreases or absent tibial shifts [88].

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 patients during walking, running, and stair climbing [128]. Branch and Hunter (1990) found that ACL deficient subjects compensate for 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 [17]. However, when performing dynamic cutting manoeuvres that stressed the braces to a greater degree, the same support was not evident [17]

Neuromuscular response to ACL rupture

Knee stability arises primarily from two restraint systems; the *passive restraint system* that is comprised of the ligaments and capsule, and the *dynamic restraint system* that consists of the neuromuscular elements [139]. The ACL primarily resists excessive anterior translation of the tibia relative to the femur [3,14,77].

In-vivo studies have identified that cruciate receptors respond to mechanical deformation or variations in ligament tensions, tibial position and movement [37,45,60,61,75]. These mechanoreceptors translate this information into neural impulses, which are in turn sent to the central nervous system to mediate the protective reflexes of the hamstrings and quadriceps in assisting dynamic joint stability [17]. Some have proposed that functional adaptations occur after injury, e.g. compensatory muscle recruitment in order to maintain joint stability [17,139].

Following ACL injury, new demands are placed on the neuromuscular components. Delayed, absent or out of phase timing sequences may reflect adaptive firing patterns. Moreover, changes in muscle amplitude and duration have been documented. This may reflect an adaptive mechanism which may help improve functional instability [17,130,139,145]. As such, the effects of muscular activity in offsetting or accentuating ACL deficiency while substantially altering knee kinematics must be considered [120].

The relationship between the quadriceps, hamstrings and calf musculature in maintaining knee joint stability in ACL deficient knees has been well documented [17,18,29,35,48,111]. The premise involves the hamstrings functioning synergistically with the ACL during quadriceps contraction to prevent anterior displacement of the tibia relative to the femur. In cadaver studies, Hirokawa *et al.* [53] reported quadriceps contraction during knee extension had a direct impact on anterior tibial translations and rotations. ACL strain increased with increasing muscle force. More *et al.* [90] reported hamstring activity decreased the amount of anterior tibial translation and internal tibial rotation during flexion. However, hamstring load had no measurable effect on quadriceps force. Anterior tibial translations significantly increased after sectioning the ACL whereas in ACL reconstructed knees, the hamstrings greatly reduced the graft load. The authors suggested that hamstring activity probably functioned synergistically with the anterior cruciate ligament to provide anterior knee stability. However, cadaver studies are limited since it excludes the compliance of living tissue and effects of active muscular contractions.

Ciccotti *et al.* [29] compared fine-wire EMG profiles of normal controls, ACL deficient patients and ACL reconstructed subjects during various activities. They reported the presence of a co-ordinated quadriceps-hamstring muscle response rather than a precise simultaneous co-activation for all groups. The hamstrings were initiated prior to or following quadriceps activity. For the ACL deficient group, despite rehabilitation, they required greater muscle activity than normal subjects but the ACL reconstructed group produced EMG profiles statistically similar to normal subjects. With a rehabilitative strengthening program for the hamstring muscles, the synergistic behaviour between these muscles and the ACL in maintaining knee joint stability has been documented [18]. Quadriceps muscle force increased anterior drawer whereas hamstring forces had the opposite effect [125].

Neuromuscular response to functional knee braces

More research is required with respect to the effect of functional bracing on muscle behaviour during strenuous physical activity [130]. By performing an electromyographic (EMG) analysis, this may provide information pertaining to the timing of muscle activity and the relative intensity of muscle activity as the muscle produces tension [137]. During a dynamic movement, neuromuscular activity is represented by continuously changing tension levels and the Linear Envelope (LE) of the EMG signal resembles the rise and fall of muscle tension [137,138].

Perhaps knee braces aid in proprioception and muscular re-education after ligament injury. The lack of supportive evidence for bracing has led investigators to speculate that subjective approval for bracing may be the result of increased proprioceptive feedback rather than any stabilizing effect of a brace [131]. It is possible that use of a knee brace might enhance proprioception and muscular re-education after ligament injury [17,18,30]. Branch *et al.* [17] defined proprioception, as "joint position sense". A decrease in proprioception is thought to contribute to the decline of the anterior cruciate deficient knee. In order for a brace to act as a proprioception mechanism, it must provide information to the brain to alter motor responses and augment dynamic stability to the joint [95]. The current belief is that knee braces may influence the afferent input and proprioception from the ACL deficient knee to the central nervous system in high loading situations, like during the one legged hop, and therefore one might expect changes in muscle firing patterns, amplitude and timing [95]. With the lack of supportive evidence that braces provide dynamic stability, a perceived improvement in knee stability may be the result of a proprioceptive feedback and not to the brace itself. It has been suggested that bracing the ACL deficient knee may require less muscular stabilisation as a result of agonist/antagonist co-contractions. Subjective approval for bracing may place athletes at greater risk of generating increased forces and theoretically increases the risk of joint damage.

Acierno *et al*. [2] reported that highly active or asymptomatic ACL deficient subjects when fitted with a brace experienced no change in muscle activity during maximal isokinetic concentric knee extensions but although extension force decreased throughout the range of the brace. Conversely, the symptomatic group exhibited increased quadriceps activity, decreased hamstring activity and displayed a minor increase in extension force from 80º to

40º of flexion. The authors suggested the brace prevented quadriceps inhibition among the symptomatic group by exerting a posteriorly directed force to the superior antero-proximal part of the tibia; thus, the brace compensated externally for the absence of the ACL. During the stance phase of side-step cutting, Branch *et al.* [17] reported ACL deficient subjects demonstrated a decrease in quadriceps and gastrocnemius activity (both in area under the curve and peak EMG activity) but an increase in medial hamstring activity when compared to controls. Bracing the deficient limb resulted in both the hamstrings and the quadriceps being reduced significantly. Németh *et al.* [95] measured EMG activity among elite level skiers with ACL injury during downhill skiing, with and without braces. Group analysis showed that without a brace, activity of the quadriceps, medial and lateral hamstring, and gastrocnemius medialis all increased during knee flexion.

With new demands being placed on the neuromuscular components of the unsupported injured knee during loading, EMG analysis during braced and non-braced conditions is required and may help determine whether neuromuscular activity is altered during stressful dynamic activity.

AIMS OF THE STUDY

This investigation focused on the *in-vivo* three-dimensional kinematic and electromyographic analysis of the normal and ACL deficient knee. Specific aims were to:

Study 1: Assessment of functional knee bracing: an in vivo three-dimensional kinematic analysis of the anterior cruciate deficient knee.
To report whether functional knee braces worn during the modified one legged jump reduced anterior tibial translations.

Study 2: Electromyographic and biomechanic analysis of anterior cruciate ligament deficiency and functional knee bracing
Examine the effect of functional knee bracing on the neuromuscular response when performing the modified one legged jump.

Study 3: A three dimensional kinematic profile of normal tibiofemoral joint motion during strenuous activity
Develop a skeletal tibiofemoral joint profile of the normal knee. Emphasis was to compare the kinematics of the ACL deficient knee against the joint profile of the normal during dynamic activity such as the modified one legged jump and the lateral cutting manoeuvre.

Study 4: Methodological concerns using intra-cortical pins to measure tibiofemoral kinematics.
Calculate the load force required from biological tissue to permanently bend the intra-cortical pin and to propose an alternative insertion approach.

Study 5: Tibiofemoral contact points relative to flexion angle measured with MRI.
Identify articular tibiofemoral bony contact points and determine whether knee flexion influenced bony contact movements at various degrees of flexion.

MATERIALS AND METHODS

SUBJECTS

In total 29 subjects were involved with this thesis (Table 1). Of these, six patients with ACL rupture and 8 normal controls participated in the invasive pin experiments. Sixteen normals were selected for the MRI study.

Table 1: Distribution of subjects from the different papers Kin = Kinematics derived from pins (Studies I and III), EMG = Neuromuscular activity (study II), Pin = Pin bending investigation (Study IV), MRI = Magnetic resonance imaging (Study V).

Study	n	Type	Description
I	6	Kin	Young males with ACL rupture and no surgical treatment.
II	6	EMG	Simultaneous EMG measurements using subjects from study I.
III	8	Kin	Young males with normal knee function and no previous injury.
IV		Pin testing	
V	16	MRI	Young adults with normal knee function and no previous injury.

Patient characteristics (I and II)

An orthopaedic surgeon examined the patients before they participated in studies I and II. None had surgical treatment prior to the study. Each had a history of functional instability that caused them to modify their activity. Deficient knees scored +2 on the Lachman's test; they were evaluated with the KT 1000 arthrometer (MEDmetric Corporation, San Diego, USA) and compared against their contralateral leg. Table 2 highlights the personal data for the ACL-deficient group.

Table 2: Means and SD of the ACL-D subjects (Studies I and II).

Subjects (ACL-D group)	A	B	C	D	E	F	Mean	SD
Age (Yrs)	29	19	19	24	22	18	21.8	± 4.2
Mass (Kg)	75	79	72	92	84	77	79.8	± 7.2
Height (cm)	176	180	175	184	180	194	181.5	± 6.9
Deficient Leg	Left	Left	Left	Right	Right	Right		
Lysholm Knee Score	72	75	75	69	74	70	72.5	± 2.6
Tegner Activity Score	4	9	6	7	6	4	6.0	± 1.9
Lachman scores	+2	+2	+2	+2	+2	+2		
Max. KT 1000 values (mm)	18	15	10.5	11	14	12.5	13.5	± 2.8
Difference between limbs (mm)	9	5	4.5	5.5	5.5	5.5	5.8	± 1.6
Randomized brace application	2nd	1st	2nd	1st	1st	2nd		

Tegner scores [127] (perceived rating of the level of activity before and after the injury) for the ACL deficient subjects' suggest a wide variation in "coping" with the injury. All modified their activity level albeit Subject B continued to play soccer. The patient reportedly wore a soft (prophylactic) brace since he felt it gave him added stability. However, his play was modified to accommodate for the injury. Three subjects were physically active to some degree after ACL rupture but unable to return to their original level of activity whereas two were not physically active after injury. Tegner Scores were reported between 7 and 8 prior to the injury.

Ethics

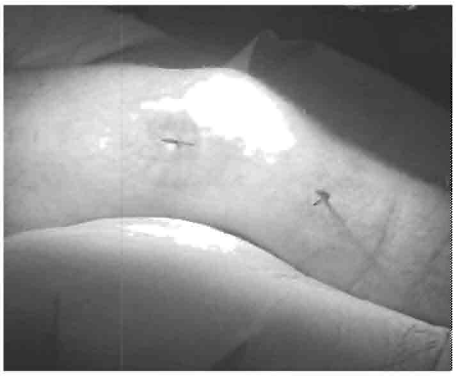
The Ethics Committee of the Karolinska Hospital approved the experimental procedures. Participants signed an informed consent form to participate in each of the studies.

DATA ACQUISITION

Bone pin protocol (I, II and III)

In studies I, II, and III, 3.2 mm Hoffmann percutaneous pins (Stryker Howmedica AB, 3 mm diameter, #5038-5-80) were inserted with the femur and tibia with the knee in slight flexion to minimise impingement problems with the Iliotibial (IT) band [89]. The femoral pin was inserted superior to the lateral femoral condyle and directed obliquely in a posterior-medial direction (Figure 3 A & B).

The tibial pin was directed into the medial aspect of the tibial shaft postero-laterally for study I. The pins were inserted well below the tibial plateau so as not to impinge with the brace during testing. A lateral insertion site was not applicable since the pin would have penetrated the tibialis anterior muscle. Conversely, for study III, more of a lateral insertion site was selected. Target clusters were then affixed to the pins. No flexion-extension impairments resulted from impingements between the iliotibial band and the femoral pin or the brace/pin interface.



A

B

Figure 3A & B: For studies I and III, the femoral and tibial femoral pins were implanted oblique angles with the limb in slight flexion.

Stereophotogrammetric radiographs (RSA) were taken with target markers affixed to the pins to identify the femoral and tibial anatomical reference points (Figure 4). As illustrated in figure 5, the femoral anatomic reference point was defined as the deepest point of the intercondylar groove. The superior aspect of the medial intercondylar eminence was identified as the tibial anatomic reference point.

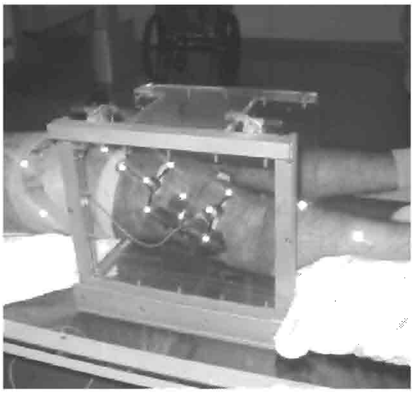
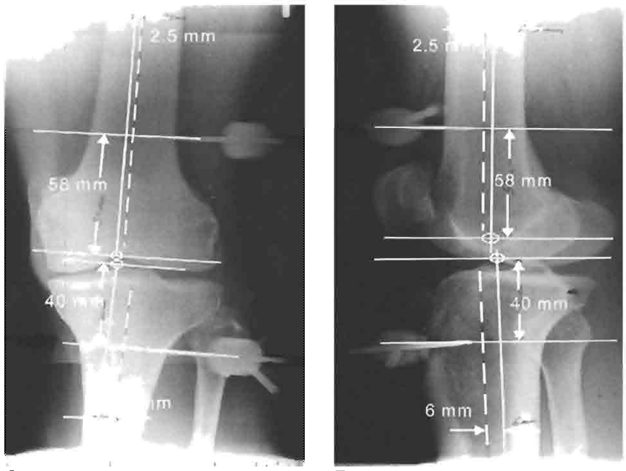


Figure 4: Set-up for stereophotogrammetric X-rays (RSA) with markers affixed to the pins (studies I and III)



A B

Figure 5A and B: A) Frontal and B) sagittal X-rays (RSA) depicting the femoral and tibial anatomical reference points (studies I and III).

Mechanical testing of the Hoffman pin (IV)

Hoffman pins (3.2 mm) underwent mechanical testing using a hydraulic load frame (MTS Systems Corp, Model 305.03, 50/100 KIP, Minneapolis, Minnesota) and digital controller (Instron 8500, Instron Corporation, Canton Massachusetts). A force transducer (Kraftaufnehmer, Hottinger Baldwin Messtechavik, Type U2B, Germany) with a load range of 0.5 kN was instrumented to the load frame. Pins were secured in steel vice at a depth of 18 mm, (representative of the insertion depth into the femur) and fully inserted so no threads were exposed. Each pin underwent one test condition. Loads were transmitted perpendicular to the pin at 15 mm or 20 mm from the insertion site at a constant rate of loading and unloading, equal to 0.02 m/s (Figure 6). To determine the resonance of the pin and triad, a marker cluster was affixed to the end of a new pin and the set-up vibrated.

Figure 6: Bone pin undergoing mechanical testing (study IV).

Kinematic MRI (V)

Recordings were made from the right knee of each subject fixed at three different angles using a customized adjustable orthosis for accurate positioning, 0° (limb supine and fully extended), 30° and 60° of flexion. Images were acquired on a 1.5 tesla (T) Sigma System

(General Electrics (GE), Milwaukee, WI). Using a 3-D GRASS pulse-sequence (Gradient Refocused Acquisition in the Steady State), images were acquired as follows: TR 24ms, TE 8ms, FA 50°, FOV 24cm, Slice thickness 1.0 mm, Number of slices 124, Interscan spacing 0.0mm, Matrix 256x192 and NEX of 1.5.

Motion recordings (I, II and III)

For study I and II, 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 synchronized 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. The cameras captured all the markers about the knee (approximately 45cm off the floor) within a $0.02m^3$ (0.49 m × 0.2 m × 0.25 m) measurement area. In study III, tibiofemoral kinematics were recorded with four ProReflex infrared cameras (Qualisys, Sävedalen, Sweden) sampling at 120 Hz within a 0.8 m^3 (1.1 m × 0.8 m × 0.9 m) volume. Motion recordings, ground reaction force and muscular activity were synchronized to collect simultaneously.

Simultaneous ground reaction force recordings (I, II and III)

Ground reaction forces were collected with a Kistler force plate (Kistler Instruments AG, Winterhur, Switzerland) to control for experimental variables. Analogue signals from the force platform were sampled at 960 Hz, A/D converted at 12 bit resolution and stored on SC/ZOOM, a dedicated signal analysis computer system (Department of Physiology, Umeå University, Sweden).

Simultaneous EMG recordings (II)

Using pre-amplified surface electrodes with a gain of 10 (IK Elektronik, Ellös, Sweden), muscular activity was recorded from the rectus femoris, semitendinosus, bicep femoris, and lateral head of gastrocnemius and later analyzed relative to the movement patterns. Each consisted of bipolar silver-silver chloride electrodes with an interelectrode distance of 20 mm center-to-center. The ground electrode (Blue sensor VL-00-S, Medicotest A/S, Denmark) was placed over the greater trochanter. Amplifier gain was set at 100 and the raw EMG signals were high pass filtered (third order) at 8 Hz and low pass filtered (eighth order Butterworth) at 800 Hz. To avoid the influence of movement artefacts, electrodes and cables were taped to the subject.

TESTING PROCEDURE

For studies I, II, and III, patients jumped for maximal horizontal distance to sufficiently stress the knee and ACL (Figure 7).

Due to the invasive nature of our protocol, hopping from and landing onto the deficient leg when affixed with pins may increase the risk of complications. Therefore, the hop was modified whereby patients jumped for maximal horizontal distance, pushing off from their sound limb and landing onto their deficient limb, within their own comfort limits.

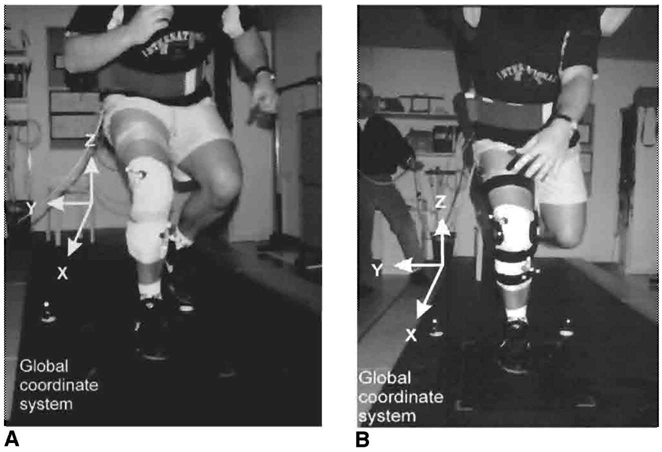


Figure 7A & B: Patients jumped for maximal horizontal distance and landed onto their deficient limb with the knee unbraced (A) and (B) braced. Subjects pushed off from the sound limb and landed with the deficient limb to provoke a subluxation (study I, II, and III).

Subjects were given several trials to perform the one legged jump to familiarize themselves with the pins and testing protocol. The longest measurement was 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. The ACL deficient patients were randomly assigned to start with either the braced or non-braced condition. The DonJoy Legend knee brace (Smith & Nephew DonJoy Inc., Carlsbad, USA) was carefully applied as prescribed by the manufacturer. No impingement occurred between the brace and Hoffman pins. Five measurement trials and two neutral standing trials were recorded for normal controls and for each brace condition for the ACL deficient group.

DATA PROCESSING AND ANALYSIS

Kinematic and kinetic analysis (I, II and III)

The kinematic data was 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). All movement patterns were referenced to a standing reference trial recorded prior to activity. The orientation of the target clusters from the first standing trial was used a reference against the second standing trial to verify that the triad did not rotate on the pin during testing.

Kinematic data for the ACL injured group and normal controls were filtered with a Butterworth 4th order, low-pass, critically damped, zero-lag filter at 6 Hz and 15 Hz respectively. The cut-off frequency was determined by running a Fourier analysis that retained 95% of the original signal (both angular and linear data) and by visual inspection. Cardan angles were employed to describe the sequence of rotations using the conventions of Kadaba [65] and Davis [31] and computed about –y, x, z axes [66]. Tibiofemoral joint motion was described using Grood and Suntay's joint co-ordinate" system [47].

Synchronization and normalization

Foot-strike was obtained from the force platform data and the corresponding frame number identified in the kinematic data. For each subject and condition, the time axes for both kinematic and kinetic data were normalized to percent. For the ACL deficient group in study I and II, the jump sequence commenced 50 ms prior to foot-strike through to knee extension and when the associated posterior shear force began to plateau (Figure 8).

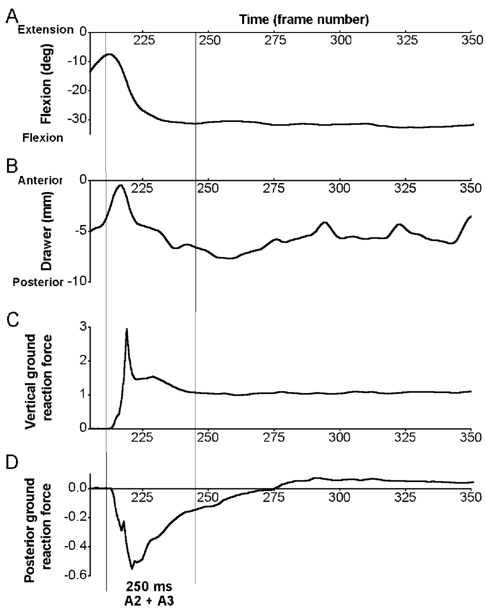


Figure 8: Data illustrating the one legged jump. **A:** Flexion/extension, **B:** Drawer, **C:** Vertical ground reaction force, **D:** Posterior shear force. Impulse is the area under the vertical ground reaction force curve and was derived within the 250 ms time interval (A2 + A3). Reprinted with permission from Elsevier Science

For normal controls, the cut cycle was normalized as follows: the time in stance (footstrike through to toe-off) was derived from the force platform data and normalized to 100%. From the kinematic data, the longest time interval that was visible within the calibrated volume prior to footstrike was calculated as 9%. Therefore, the cut cycle included the approach phase and stance phase and normalized to 100%. Average movement patterns were derived using trials collected for each of the four subjects during unbraced and braced testing and for each of the normal controls during the jump and cutting manoeuvres.

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.

Ground reaction force data were used as a control across trials and between brace conditions. Vertical and posterior shear forces were scaled to body weight (including the brace when applicable) and time normalized following the same criteria used to normalize the MacReflex and ProReflex (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 differences in the landings onto the force platform. It was arbitrarily defined that peak vertical 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. Additionally, the impulse created upon landing was calculated and examined. Impulse is the product of force over the time interval with which the force acts. During landing, impulse is equal to the area under the respective vertical force-time curve.

Accuracy and validation

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 [86]. To quantify the accuracy of the experimental procedure, separate motion recordings were performed using a skeletal tibiofemoral reference standard complete with bone pins and target markers. The identical experimental protocol, data reduction and analysis were followed in order to derive the 3D kinematics of the reference standard. The reference standard was carefully moved into each position so that no secondary rotations or translations occurred. Angular rotations were measured using a hand held goniometer and linear displacements were measured using a vernier calliper about the anatomical reference points. Prior to each predetermined displacement, a standing reference trial was recorded.

This mechanical model combined with the straightforward procedure provided validation in terms of clinically relevant parameters.

The accuracy and consistency of the 3D angular and linear estimates were generally well reflected. Rotations less than 3° for flexion and less 2° for non-primary rotations were observed. These errors may be considered large but is likely the result of inherent inaccuracies using the hand help goniometer or procedural artefacts due to small discrepancies in its alignment. Accuracy of the linear estimates was found to be less than 2 mm for all translations. Errors would be have been smaller if a calibrated mechanical representation of the human knee [32] was employed.

EMG analysis (II)

The EMG signals were sampled at 960 Hz and A/D converted at 12 bit resolution (SC/ZOOM, Department of Physiology, Umeå University, Sweden). Data was high-pass filtered (Butterworth 4th order, critically damped, zero lag filter) at 20 Hz to remove the remaining artefacts. The raw EMG was full wave rectified and low pass filtered with a cut-off frequency of 10 Hz producing an analogue linear envelope (Bioproc Version 1.65c, School of Human Kinetics, University of Ottawa, Canada). The conservative cut-off frequency was used in order to keep some of the high-frequency information while preserving the common muscle firing patterns.

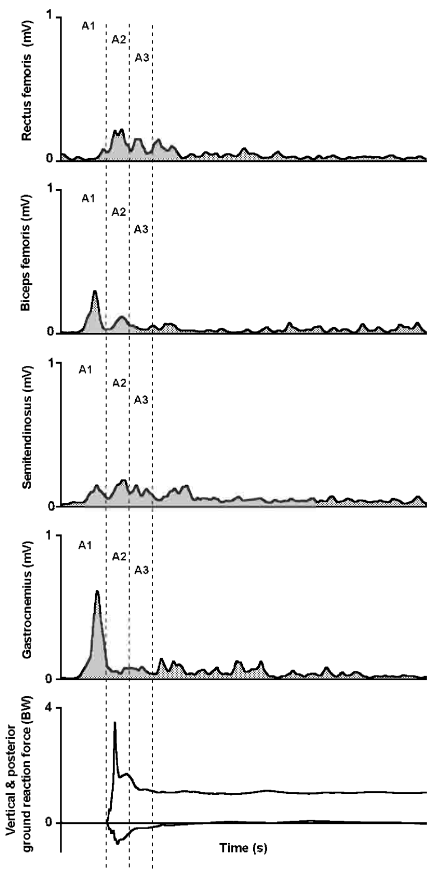


Figure 9: A single trial that illustrates the linear envelope (cut-off 10Hz) and the IEMG for each muscle at the determined time intervals. A1: 250 ms prior to footstrike, A2 & A3 are consecutive 125 ms time intervals following footstrike. Reprinted with permission from Elsevier Science.

Initial contact with the force platform was noted to co-ordinate EMG and ground reaction force data. For each subject and brace condition, EMG and force data were marked to a specific time period. To estimate muscular activity, the linear envelope was integrated producing a singular arbitrary value that represents the area under the curve. Integrated EMG (IEMG) was calculated for three time windows: 250 ms preceding footstrike (FS) and two 125 ms intervals following FS (Figure 9). Therefore, analysis focussed on the transition between non-weightbearing and weightbearing phases separately. Time intervals no less than 125 ms were selected in order to overcome the issue electromechanical delay. Of interest were anticipatory changes prior to FS and the 125 ms time interval after the foot contacted the force plate to peak Fy. Each trial was considered an independent sample within each bracing condition. Averages were derived for each subject's muscle for both bracing conditions. Subjects served as their own control with analysis focussed on differences in IEMG magnitudes between bracing conditions.

Statistics (II)

Since subjects served as their own control, the Friedman's ANOVA by rank for *k* related samples was used to compare within subject measures in both brace settings. The Friedman test is the non-parametric equivalent of a standard repeated-measures analysis of variance applied to ranks rather than raw scores [55,114]. The point for using a matched test was to control for experimental variability between subjects, thus increasing the power of the test. For each subject, matching was achieved by comparing the means of repeated measures for each muscle and time interval for both non-braced and braced conditions. If the ranks between conditions were very different, the notion that differences are coincidences of random sampling was rejected. Differences were considered significant when the probability of an α type error was < 0.05. Emphasis was to highlight significance to this particular group with respect associated anterior tibial translation.

MRI analysis (V)

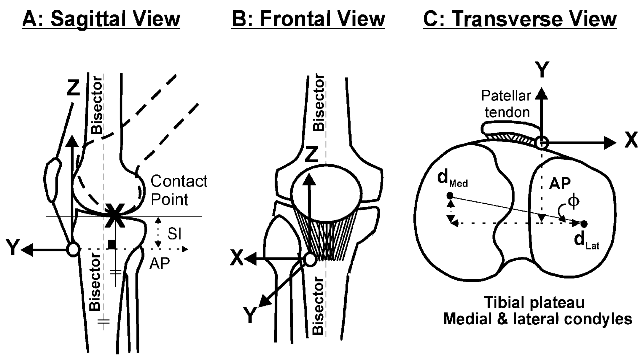
All measurements were made by the same radiologist. Repeated independent measurements showed an intraobserver variation of less than 0.5 mm. The centre of the contact areas between the femur and tibia on the medial and lateral condyles were analysed from slice images in the sagittal and frontal planes.

The most anterior and the most posterior (sagittal plane) co-ordinates as well as the most medial and the most lateral (frontal plane) co-ordinates were marked for both the medial and lateral condyles (Figure 10). These four co-ordinates determined a contact area on the medial and lateral compartments of the knee. From these contact areas, the centroid points in all three planes were calculated. This gave one contact point for the medial and lateral knee compartment respectively.

A B

Figure 10A & B: Example of co-ordinate extraction for the two frontal points (A) and sagittal points (B) of the lateral contact of area between the tibia and femur at 30° knee flexion (Study V).

We identified the anatomical co-ordinate system as the most lateral and distal site of the patellar tendon insertion (Figure 11). Movement of the contact point was reported as a change in position relative to the anatomical reference point. The position of the contact point with respect to the reference frame was determined as follows: In the sagittal view, the AP line of action was perpendicular to the line of action of the contact point (Figure 11a). The location was calculated as the distance from the origin to the line of action of the contact point.



SI: Contact point relative to the anatomical reference frame along the longitudinal axis.
AP: Medial and lateral contact point relative to the anatomical reference frame along the anteroposterior axis.
d_{Med}: Perpendicular distance of the medial contact point relative to the anteroposterior axis.
d_{Lat}: Perpendicular distance of the lateral contact point relative to the anteroposterior axis.
ϕ: Rotational angle about the longitudinal axis, [$\phi = \mathrm{Tan}^{-1}$ $(d_{Med} + d_{Lat}) / (AP_{Lat} - AP_{Med})$]

Figure 11: Position of the contact point with respect to the anatomical reference frame (Study V). Reprinted with permission from Elsevier Science.

Along the longitudinal axis, the position was calculated from the contact point to the orthogonal anteroposterior (A/P) line of action. In the transverse plane (Figure 11c), the medial/lateral (M/L) line of action was perpendicular to the A/P line of action. The medial and lateral contact points were calculated as the distance from the A/P line of action to the contact point along the M/L line of action. From both the A/P and M/L co-ordinates rotations were calculated using the Pythagorean theorem.

Only when the leg was supine was the MRI co-ordinate system aligned with the anatomical reference frame. To correct this when the knee was flexed, the lines of action derived from MRI co-ordinates at 30º and 60º were transformed mathematically and aligned with the anatomical reference frame (Figure 12).

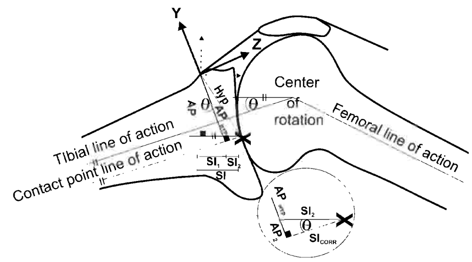


Figure 12: Knee model used to derive the correct contact position for 30° and 60° (Study V). Reprinted with permission from Elsevier Science.

RESULTS

TIBIOFEMORAL JOINT MOTION DERIVED FROM BONE PINS (I AND III)

No subjects experienced significant discomfort, all reported they could move their knees freely and their ability to jump was unaffected by the pins. Of the 14 subjects, five subjects generated enough force to bend the femoral pin during knee flexion, the result of an interaction with the soft tissue and musculature.

Two subjects were excluded in study I. Subject E was excluded as a result of significant marker dropout in the kinematic data. Subject F was excluded due to the femoral pin bent during flexion. Two normal controls were removed from study III. One similarly bent the femoral pin bent whereas the second exclusion was because transformation from the global coordinate system to the anatomical coordinate system was impossible. One subject bent the pin prior to RSA x-rays being taken and measurement recordings. The standing reference trial recorded prior to activity was matched against the second standing reference trial to verify the orientation of the target clusters were similar. Moreover, kinematic data was checked to ensure the triads did not rotate on the pin during testing. Therefore, data for this subject was included in the analysis.

Braced vs. non-braced (I)

Figure 13 illustrates flexion-extension and anterior-posterior drawer patterns between bracing conditions and across subjects. Intra-subject anteroposterior drawer curves were similar in shape between and fairly similar across subjects, i.e. kinematics were repeatable. The tibia displaced anteriorly with respect to the femur from footstrike to approximately peak Fy. Thereafter the tibia was drawn posteriorly during flexion. Two subjects exhibited greater flexion RoM when the knee was braced. Following peak flexion, three subjects stabilized the knee and remained in flexion overall whereas one subject returned to full extension. Offsets were evident between bracing conditions.

Table 3 depicts the differences in angular and linear RoM between non-braced and braced conditions. Negative values indicate reductions in amplitudes during bracing. Differences in anterior displacements between unbraced and braced conditions were small. During bracing, anterior displacements remained unchanged for one subject; two subjects demonstrated small reductions while anterior displacements were larger for the remaining subject.

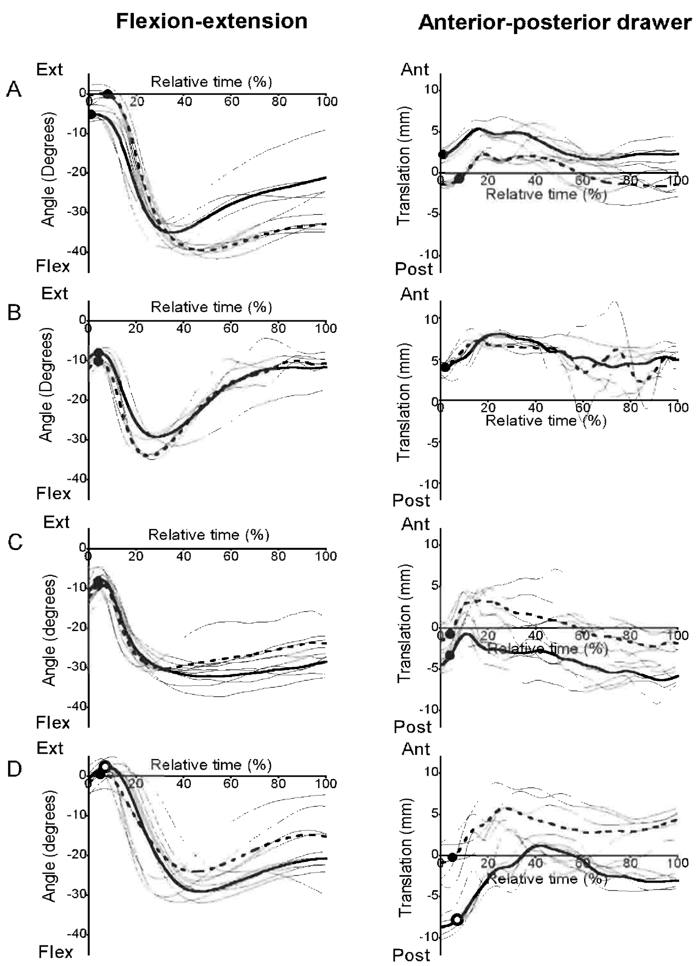


Figure 13: Flexion/extension and anterior translations derived from skeletal markers. Means of are displayed in bold. The bold solid line represent the non-braced kinematics, the bold dashed line represent braced kinematics. Foot-strike is identified as an open circle derived from force platform. Closed circle derived from kinematics.

Table 3:Differences in angular and linear displacements (RoM) as a result of bracing.

Subject	Trials	Angular (°)			Linear (mm)		
		Flex-ext	Abb-add	Int-ext	Med-lat	Ant-pos	Dis-comp
A	n = 5	10.0	3.2	- 1.2	1.6	- 0.7	- 5.1
B	n = 3	2.6	0.7	- 0.9	- 0.4	- 1.1	- 2.3
C	n = 5	- 2.9	- 1.1	0.8	0.3	1.3	- 1.9
D	n = 5	- 6.9	- 3.0	- 5.0	- 3.0	- 3.1	- 2.4

i) A negative value indicates that flexion, abduction and external rotation
ii) A negative value indicates a medial shift, posterior displacement and distraction.

Table 3 depicts the differences in angular and linear RoM between non-braced and braced conditions. Negative values indicate reductions in amplitudes during bracing. Differences in anterior displacements between unbraced and braced conditions were small. During bracing, anterior displacements remained unchanged for one subject; two subjects demonstrated small reductions while anterior displacements were larger for the remaining subject.

Normal controls (III)

Figure 14 illustrates normal joint kinematics for each subject during the hop and side cut respectively. Flexion patterns were similar in shape between subjects and across tasks although differences in RoM were noted. Table 4 depicts flexion and translation RoM during three time intervals; during flexion, peak flexion and extension. For the jump, all subjects demonstrated anterior tibial displacements with respect to the femur although when peak occurred varied. Thereafter the tibial reference frame was drawn posteriorly.

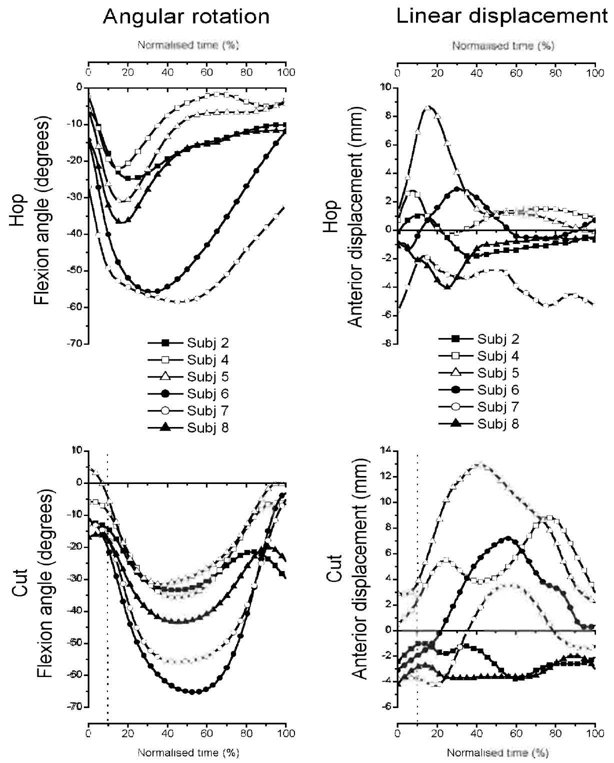


Figure 14: Subjects tibiofemoral flexion-extension and anteroposterior displacements normalized to time for the one legged jump and the lateral side cut.

For the cut, heel-strike is depicted as the vertical line at 10% of the cycle. Anteroposterior displacements showed poor agreement in the shape of the patterns across subjects. When peak occurred and RoM varied considerably (Table 4). As observed for the jump, overall tibial translations were anterior up to approximately peak Fy or peak flexion or about 50% of the cycle. The tibia displaced posteriorly thereafter.

Table 4: Means of angular and linear ranges of motion for the jump and cut.

		One legged jump					Lateral side cut				
		Angular °		Linear (mm)			Angular °		Linear (mm)		
Subj	n	Flex	Ext	33.3% flex	66.6% mid	100% ext	Flex	Ext	33.3% flex	66.6% mid	100% ext
2	5	-18.8	14.7	1.2	-0.7	1.3	-20.8	11.9	1.4	-2.8	1.7
4	5	-19.8	20.3	2.1	-3.0	1.7	-29.9	29.7	5.5	-1.7	-5.8
5	5	-41.4	43.9	4.2	-3.5	1.3	-53.6	62.1	10.3	-6.9	
6	5	-31.4	26.3	3.8	-3.5		-43.0	50.5	-0.6	7.8	-4.9
7	4	-22.0	25.0	-3.1	3.5		-27.2	23.2	1.5	-1.0	1.0
8	5	-25.0	26.8	7.3	-7.7	- 1.2	-26.6	31.3	10.1	-10.7	

i) Negative value indicates knee flexion and posterior displacements of the tibia relative to the femur.

Kinetics (I, II, III)

For the ACL injured group in study I and II, mean intra-subject peak vertical force at foot-strike, peak posterior shear force and impulse were generally consistent between unsupported and braced conditions (Table 5). Peak forces varied across subjects since subjects jumped within their own comfort limits.

Table 5. Mean peak vertical force and mean peak posterior ground reaction force (± SD) normalized to bodyweight and mass of the brace across subjects and conditions. The impulse for peak vertical force was derived using a 250 ms interval following heel-strike. Values expressed relative to bodyweight.

		Vertical (Fy)		Impulse for (Fy)		Posterior shear (Fx)	
Subject	Trials	Unbraced N/kg	Braced N/kg	Unbraced N/kg/s	Braced N/kg/s	Unbraced N/kg	Braced N/kg
A	n = 5	2.9 ± 0.4	2.6 ± 0.1	0.39 ± 0.02	0.38 ± 0.01	-1.3 ± 0.2	-1.1 ± 0.1
B	n = 3	2.2 ± 0.3	2.4 ± 0.1	0.36 ± 0.03	0.39 ± 0.01	-0.6 ± 0.2	-0.9 ± 0.1
C	n = 5	3.4 ± 0.36	2.6 ± 0.6	0.35 ± 0.01	0.36 ± 0.02	-0.7 ± 0.1	-0.6 ± 0.1
D	n = 5	n/a	2.9 ± 0.3	n/a	0.37 ± 0.03	n/a	-1.1 ± 0.0
E	n = 5	2.9 ± 0.6	2.0 ± 0.1	0.46 ± 0.02	0.39 ± 0.02	-1.0 ± 0.2	-1.0 ± 0.1

n/a: data not available

The braced leg showed lower peak vertical forces that exceeded the 0.5 time bodyweight criteria for two subjects. Overall, shear forces were consistent between conditions. With the knee supported, no changes in impulse magnitudes were observed for subjects B and C whereas one demonstrated a slight reduction. This coupled with peak vertical and peak posterior shear force indicates that jumps onto the force platform were similar between brace conditions. Any change in skeletal kinematics cannot be attributed to differences in landings but rather to the brace itself.

Although the data recording system failed to store ground reaction force data for subject D, angular data was used to determine whether jumping styles were similar between conditions.

For the normal controls, the means and (SD) of peak vertical force (Fy) are depicted in Table 6. For the hop, the single peak represents the force at landing. The two peaks for the cut represent footstrike and toe-off during stance. The consistency in GRF data indicates that jumps onto the force platform were similar.

Table 6: Means (SD) of peak vertical ground reaction force normalized to bodyweight and total stance time across subjects and conditions.

Subject	Trials	Jump	Cut		
		Peak at FS	Peak at FS	Peak at TO	Time in Stance (ms)
2	**n=5**	1.48 (0.21)	1.37 (0.05)	1.28 (0.11)	489 (30)
4	**n=5**	1.93 (0.08)	1.60 (0.11)	1.41 (0.11)	429 (29)
5	**n=5**	1.90 (0.16)	1.93 (0.12)	1.60 (0.09)	479 (44)
6	**n=5**	1.58 (0.11)	1.57 (0.12)	1.57 (0.18)	466 (25)
*** 7**	**n=3/4**	1.46 (0.21)	1.34 (0.07)	1.27 (0.05)	542 (29)
8	**n=5**	1.93 (0.11)	1.58 (0.29)	1.46 (0.13)	522 (124)

* Trials for subject 7: n = 3 for the cut and n =4 for the hop.

When GRF data was considered in combination with the results from the brace study, peak vertical force values for normal controls did not exceed 2 times bodyweight whereas the ACL deficient group experienced up to 3.5 times bodyweight. Anterior tibial displacements for the ACL deficient knee when unsupported were on average 4.4 mm (range 2.2 to 8.8 mm) and 3.6 mm (range 2.4 to 5.7 mm) with the knee braced. For the normal control group, displacements were 3.7 mm (range 1.2 to 7.3 mm) for the jump and 6.1 mm (range 1.4 to 10.3 mm) for the cut respectively. The decline in peak force magnitudes diminished anterior displacements for the jump yet sizeable anterior displacements of 6.1 mm (range 1.4 to 10.3 mm) were observed for the cut. Similar anterior translation magnitudes were observed for the non-braced and braced ACL deficient subjects and normal controls.

EMG (II)

Subject A was excluded due to dropout of the EMG signal in study II. The results from the Friedman's ANOVA by rank indicated that semitendinosus activity decreased by 17% during A1, the 250 ms interval prior to footstrike ($P<0.05$). Biceps femoris significantly decreased 44% during A2, the 125 ms interval following footstrike ($P<0.05$). Conversely, rectus femoris activity significantly increased 21% with the brace during A2, ($P<0.05$). However, this does not take into account timing and co-ordination. Figure 15 illustrates the group means and standard deviations for each muscle at each time interval. Significance is indicated by a (*) and the percentage of difference.

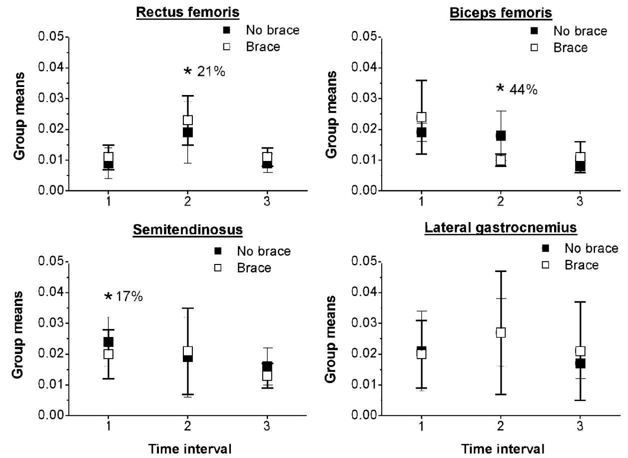


Figure 15: Group means and standard deviations for each muscle at each time interval. Significance is indicated by a (*) and a percentage of difference [105]. Reprinted with permission from Elsevier Science.

In Table 7, the change in IEMG for each muscle and time interval between non-braced and braced conditions is presented along with the patient's corresponding kinematic data [106], solely for the period A2. Only three subjects are identified since their EMG and associated kinematic data were available. In interpreting the data in Table 7, a negative value with respect to EMG data indicates reduced activity during bracing. Additionally, a negative value for anterior displacement magnitudes indicates a reduction in anterior drawer when the knee was braced.

Table 7: The difference in the IEMG between the brace conditions during A2 for each subject and muscle and matched against the change in anterior tibial displacements (mm).

	Difference in Muscle IEMG				Anterior Tibial Drawer		Δ
Subject	R. Fem mV • s	B.Fem mV • s	Semi mV • s	Gastroc mV • s	No brace mm	Brace mm	Drawer mm
B	‡ 0.007	† - 0.004	0.002	- 0.019	3.5	2.4	-1.1
C	‡ 0.005	† - 0.011	0.004	0.000	2.2	3.5	1.3
D	‡ 0.002	† - 0.001	-0.009	0.026	8.8	5.7	-3.1

Positive value represents an increase in the area during bracing
† Significant decrease in IEMG between bracing conditions for time interval A2 ($P < 0.05$)
‡ Significant increase in IEMG between bracing conditions for time interval A2 ($P < 0.05$)

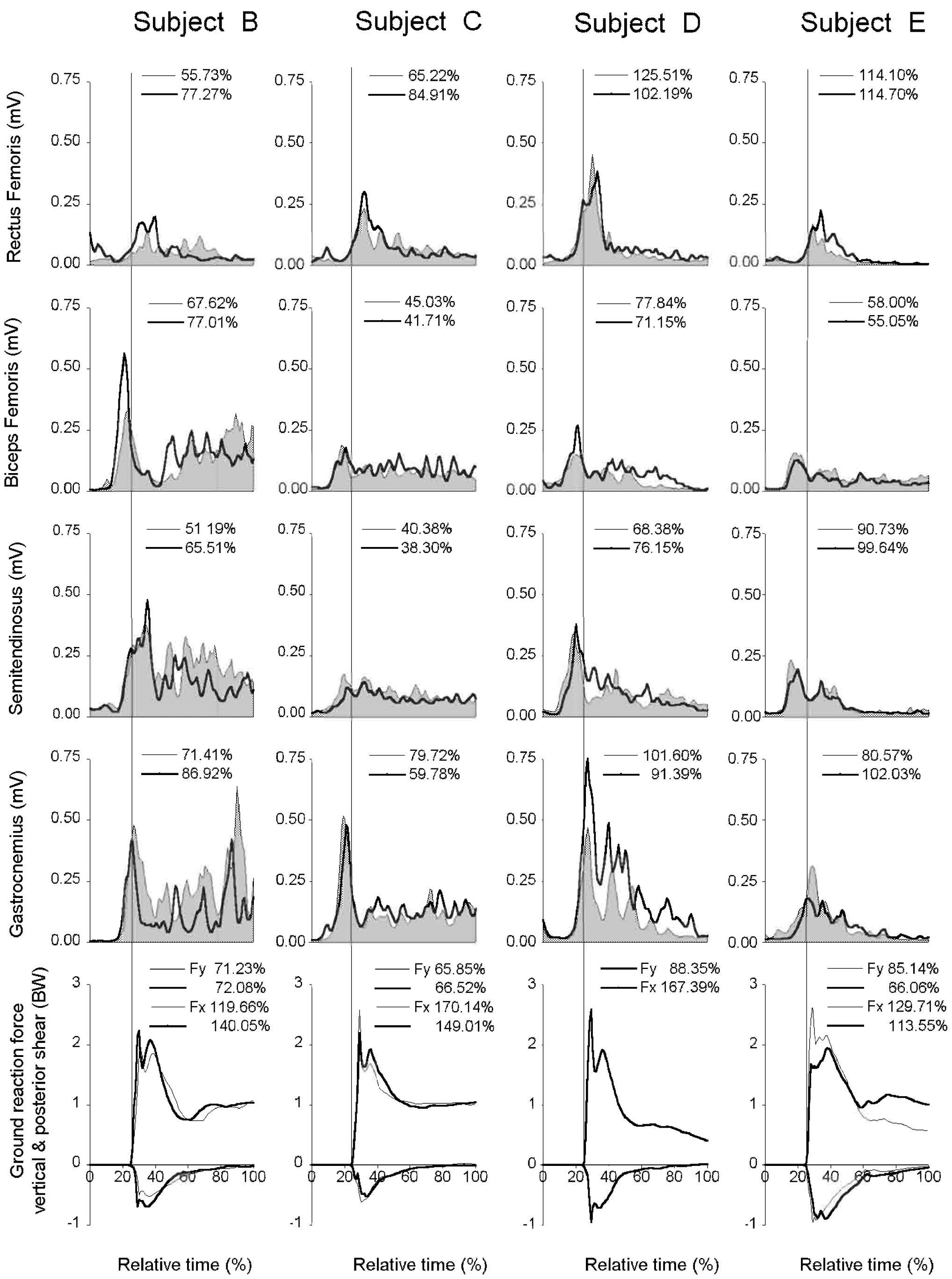


Figure 16: Ensemble average data for the rectus femoris, biceps femoris, semitendinosus, gastrocnemius and ground reaction forces. The vertical line identifies footstrike. Means are displayed as a thin solid shaded line for the non-braced condition; the solid bold line represents the braced conditions. Coefficient of variation is expressed as a percent.

Figure 16 displays descriptive mean normalized linear envelope EMG recordings with respective ground reaction forces for each subject. Except for the bottom traces (GRF), a great deal of diversity is evident in the EMG traces rather than to the homogeneity. Thus it is difficult in drawing any definite conclusion. Looking at the Gastrocnemius, peak of the burst of initial activity coincides with impact of the landing foot (Subject B), precedes impact (Subject C), arrives after impact (Subject D) and is terribly muted although coincides with peak vertical force (Subject E). The same for the semitendinosus; delayed burst (B), flat response (C), preceding impact (D), double humped but otherwise relatively silent (E).

However, within-subject EMG patterns were fairly similar between bracing conditions although temporal changes in peak activity were noticeable as a result of bracing. As observed from the kinematic data, overall the tibia displaced anteriorly immediately upon footstrike. Peak rectus femoris occurred following footstrike. Onset and peak biceps femoris activity preceded footstrike. With respect to changes in the timing of peak EMG, bracing resulted in delaying peak rectus femoris activity for subjects D and E and delayed peak gastrocnemius and semitendinosus activity for subject E. Other small differences were in timing were noted albeit negligible.

CALCULATION OF THE BONE PINS BENDING STRESS (IV)

An example of the hysteresis curve is depicted in Figure 17 that illustrates the pin exhibited plastic behaviour with no sharp yield point. It was therefore decided that bending occurred after deflections of 0.4 mm. This falls within the accuracy of roentgen-stereophotogrammetric x-ray (RSA) and MacReflex system (< 0.6° and < 0.4 mm for a 0.01625 m^3 volume) [106]. Loads of 150 N and 100 N applied at 15 mm and 20 mm respectively produced deflections larger than 0.4mm (Table 8A). Cyclic loading produced a plastic deformation of 0.25 mm in the initial loading phase but with repeated trials the pin experienced elastic deformation by returning to its original shape. Table 8B highlights deflection magnitudes for the repeated measures. A clear resonance was found at about 90 Hz. A simple calculation using the measured stiffness and the mass of the tripod (15 g) gave a somewhat lower natural frequency.

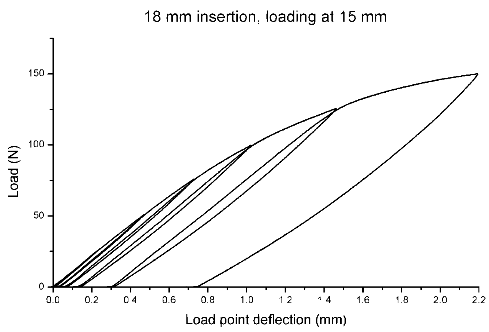


Figure 17: Loads required to deform the pin. Load applied 15.0 mm from where the pin was inserted.

Table 8: A) Summary of calculated and measured deflections during incremental loading required to permanently deform the Hoffman pin. **B)** Summary of deflections with repeated loading at 125 N.

A)

Pin	Pin Depth (mm)	Load point (mm)	Measured deflection (mm)			
			50 N	100 N	150 N	200 N
1	18	15	0.01	0.12	0.73	
2	Fixed	15	0.00	0.01	0.03	0.06
3	18	20	0.08	0.41	1.63	2.56
4	Fixed	20	0.02	0.04	0.1	0.23

B) Cyclic loads 125 N			Measured Deflection (mm)				
Pin	Pin Depth (mm)	Load point (mm)	trial				
			1	2	3	4	5
5	18	15	0.25	0.27	0.28	0.29	0.30

MRI (V)

Figure 18 illustrates the mean of subjects tibiofemoral contact points for each plane at 0°, 30°, and 60° along with associated displacements. The predominant motion was the posterior displacement within both the medial and lateral tibiofemoral compartments. Overall, the contact points on the lateral tibial condyle were positioned further posteriorly with slightly greater displacement magnitudes. Comparing movements, contact motion progressed posteriorly 12.5 mm vs. 13.8 mm, inferiorly 6.2 mm vs. 9.7 mm and laterally 0.6 mm vs. 1.4 mm for the medial and lateral condyles respectively. The predominant posterior displacement suggests femoral rolling. With increased flexion to 60°, the contact points moved anteriorly 1.0 mm vs. 6.4 mm, superiorly 2.4 mm vs. 4.7 mm and continued laterally 2.5 mm vs. 2.4 mm for the medial and lateral condyles

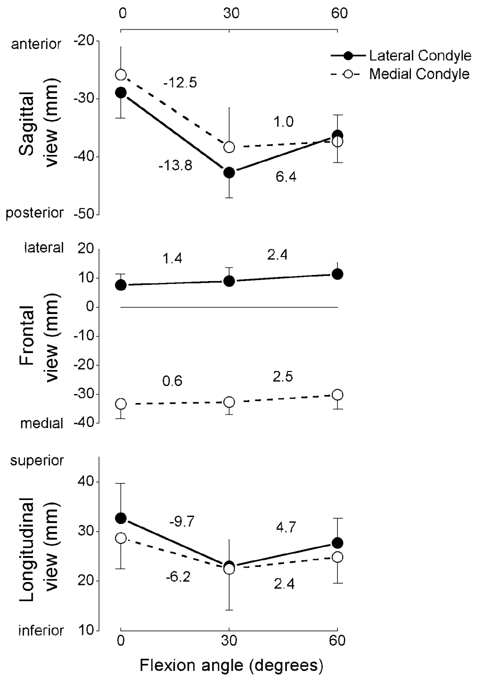


Figure 18: Mean tibiofemoral contact positions with standard deviations at 0°, 30°,and 60°. Displacement magnitudes are identified between 0°- 30° and 30° - 60° [143]. Reprinted with permission from Elsevier Science.

respectively. Only the lateral contact point displaced anteriorly, indicative that motion along medial femoral condyle was almost pure sliding.

Figure 19 illustrates the relative position of the contact points on the medial and lateral condyles respectively at 0º, 30º and 60º of flexion. Between 0° to 30° of knee flexion, the tibia externally rotated about 3° whereas from 30° to 60°, the tibia internally rotated about 5°.

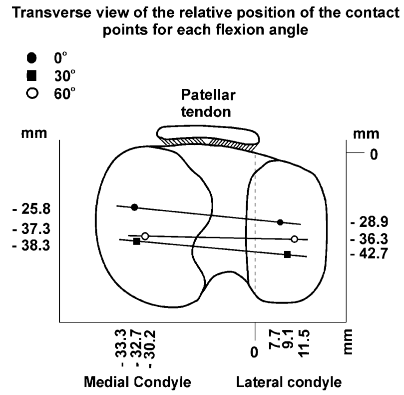


Figure 19: Transverse view of the tibial plateau showing relative tibiofemoral contact point movement on the medial and lateral condyles with respect to the anatomical reference point for 0°, 30° and 60° [143]. Reprinted with permission from Elsevier Science.

DISCUSSION

Much has been reported with respect to ACL deficiency and the effect of functional bracing on anterior tibial displacements. However, variations in experimental designs (surrogate limbs [82], cadavers [4,54,141], humans [62]) brace design, and joint loading conditions have led to conflicting interpretations thereby making comparisons between investigations impossible. [140]

Therefore, the overall aim of the present work was to investigate the *in-vivo* three-dimensional kinematics of the normal and ACL deficient knee, *primarily* the position and orientation of the tibia relative to the femur. Specific aims were to employ similar experimental designs in order to investigate whether a generic functional knee brace (i.e. DonJoy Legend) reduced abnormal anterior tibial displacements to within normal and to what extent they influenced electromyographic changes in the knee's musculature. An additional aim was to develop a non-invasive clinical tool using MRI to locate and describe 3D bony contact movements on the medial and lateral condyle. Through incremental positioning of the segment, normative baseline data can be compared against the pathological knee, e.g. ACL or meniscal injury, in order to evaluate bony contact position and the effect on joint degeneration and stability.

GENERAL LIMITATIONS

Invasive studies similar the current investigation has several limitations. First and foremost is the difficulty in obtaining the sufficient number of subjects to reach 80% statistical power. Additionally, power calculations are almost impossible since the variance has not been documented in the literature. To illustrate this, Table 9 highlights the limited number of studies and subjects that have participated in human movement studies using intracotrtical pins to describe tibiofemoral displacements.

Table 9: Summary outlining the different studies using intracortical pins to study human motion in-vivo and the number of subjects included in the research.

	Lafortune et al. [77]	McClay [89]	Reinschmidt et al. [110]	Fuller et al. [41]	Ishii et al. [58]
Subjects	5	4	* 5	1	5

* Two subjects were excluded as a result of femoral pin complications.

The small samples sizes are likely the result that patients and subjects are unwilling to participate in invasive procedures that require elective and unnecessary surgery. Furthermore, protocols such as the one undertaken for this thesis requires a huge endeavor in coordinating the operating theatre, x-ray department for RSA determination and access to laboratory facilities. Although five subjects have been generally the norm for this type of study, fourteen subjects participated in our experiments of which ten yielded useable data.

To date this is the only source of direct measurements available in the literature but these studies have contributed significantly in the understanding of skeletal tibiofemoral kinematics. However, these studies have been restricted to semi-static activities, or walking and light running. In contrast, our investigations however have challenged the knee with more strenuous and dynamic loading conditions, i.e. when landing from the one legged jump and during cutting. This is important because internal loads created by muscle contractions and compressive loading may influence displacements [11]. Yet when dealing with a limited number subjects as in our investigations, the results should be interpreted with care and treated more like baseline data that can be compared and verified in future studies. Moreover, the findings may be used to produce experimentally based hypotheses for future study designs.

TEST PROTOCOL

A common knee function test used after ACL injury is the *One-Legged Hop* for maximal horizontal distance, which requires persons to hop for maximal horizontal distance and land onto the same limb. It is generally used because it tends to stress the knee antero-laterally and it is a manoeuvre associated with the giving way phenomena during activity [42,43]. Due to the invasive nature of our protocol, hopping from and landing onto the deficient leg when affixed with pins may increase the risk of complications. Therefore, the hop was modified whereby patients jumped for maximal horizontal distance, pushing off from their sound limb and landing onto their deficient limb, within their own comfort limits.

Testing was done during a single experiment session. Subjects were given several trials to perform the one legged jump to familiarize themselves with the pins and testing protocol. Their longest measurement was marked on the floor to determine the proper take off distance to the force platform. Because the subject jumped from the same location, peak vertical force was used as a control to ensure the jumps were consistent across trials and brace conditions.

The one legged jump requires very motivated subjects. Due to the invasiveness of the protocol and since subject's jumped onto their deficient limb, jumps were within the patient's comfort limits. This may not have been enough to yield differences between test conditions. The knee can experience peak vertical forces of up to eight times bodyweight during high dynamic activity yet patients only reached about four times bodyweight. Unlike Gauffin et al., [44], the jump distance ratio (jump quotient) between the injured and non-injured limb was not calculated. It was expected that patients would be unable to produce similar lengths. Moreover, a symmetric test would not necessarily reflect adequate knee function when the knee was supported. Therefore, emphasis was on intra-subject comparisons in skeletal tibiofemoral joint motion between braced and non-braced conditions.

The DonJoy Legend knee brace was selected because of its off-the-shelf market availability since custom designed functional braces are inaccessible to many patients. Designed for moderate to severe ACL instabilities, the manufacturer claims the Legends' "Four Points of Leverage" creates a net differential posterior pre-load on the tibia that reduces knee instability and ACL strain (and therefore prevents anterior tibial translation). Beynnon *et al.* [13] reported the four points of leverage system reduced ACL strain values during weightbearing and non-weightbearing conditions when compared to unbraced knees.

Patients were selected because of their desire to return to sports activity, without any limitations. Typically classified as non-copers [36], these subjects were unable to return to their premorbid level of sports activity because of repeated episodes of giving way. None had participated in a rehabilitative thigh strength-training program but all were scheduled for ACL reconstruction. The testing protocol demanded that subjects jump for maximal horizontal distance and land onto their deficient limb. The activity was selected so the knee would be loaded to near maximal physiological values. In order to control for experimental variables, maximal horizontal distance was measured without the brace and the same distance was used during bracing. This was to control for ground reaction forces between bracing conditions. The experimental design was set up such that if differences in tibiofemoral kinematics were evident, it can be attributed to the brace rather than differences in jumping styles.

METHODOLOGICAL CONSIDERATIONS

Bone pins

Invasive markers implanted into the tibia and femur is the most accurate means to directly measure skeletal motion and may provide a more sensitive measure of the differences between brace conditions. While this method may improve accuracy, it also has some limitations. To achieve the accuracy, bone pins and markers must be rigidly attached to the bone. To account for femoral pin impingement with the iliotibial band, the knee was flexed prior to its implantation [89]. Unlike previous studies, the femoral pin was inserted superior and anterior to the lateral femoral condyle and penetrated the bone in a postero-medial direction. This has the effect of placing the Iliotibial band more posterior and should reduce impingement. It was the opinion of the co-authors that penetrating the iliotibial band was not an option as in earlier studies [110].

Ideally, subject's X-rays should have been with the subject standing rather than supine but this was not possible. Instead, X-rays were taken with the subject supine on the table with the knee unloaded. As such, transforming from the global co-ordinate system (MacReflex and ProReflex data) to the anatomical coordinate system (RSA data) was based on a standing reference trial employing an anatomical model using the RSA co-ordinate system when the knee was not loaded. This may have created artificially higher compression values and displacement magnitudes.

Previous bone pin studies have reported walking and running styles remained unaffected when using pins to record tibiofemoral joint motion in-vivo [89,107,110]. However, difficulties associated with the femoral pin have been reported [77,108,109,110]. During ballistic dynamic activity such as the modified one legged jump and lateral side cut, there is always the risk of the pins bending. In all, five of the fourteen subjects bent the 3.2 mm Hoffmann pin, the result of an interaction with the surrounding soft tissue and musculature. This may have resulted because the longitudinal incisions were too small. Deformation likely occurred as a result of the threaded portion of the Hoffman pin remaining exposed to the surrounding tissue, which is weaker due to a reduced cross sectional area.

EMG

Since EMG electrodes were applied following the surgery, no maximal voluntary isometric contractions were collected. Performing maximal voluntary isometric contractions with the bone pins implanted into the femur and tibia increased the risk of the pins bending or muscular complications. As such, inter-subject comparisons were not possible. The major disadvantage lies in the measurement scales. This provides no information on the degree of muscular activation upon landing. Knowledge of the proportion of a subject's muscle capacity required to perform a task may be important.

However, in order to estimate muscular activity the linear envelope of the EMG was integrated. This produced a singular arbitrary value for each time period that represents the area under the curve (IEMG). This enabled intra-subject comparisons that were sufficient for this study.

MRI

Tibiofemoral contact areas and pressure distributions have been measured directly on cadaveric specimens employing various invasive techniques [10,40,49,52,56,57,70,72]. The results are limited as they were obtained under non-physiological loading conditions. Pressure sensitive film has also been successfully applied *in-vivo* [112] however the articular surface contact is disturbed and changes the contact mechanics of the joint so that the true contact pressures can never be measured experimentally [144]. Whereas intracortical pins inserted into the tibia and femur are now being used to measure tibiofemoral joint motion, the contact points cannot be measured with this technique.

Therefore, kinematic MRI is a non-invasive easily accessible method for differentiating bony contact movements between the medial and lateral condyle. The primary advantage is direct and accurate registration of data. This methodology substantially improves the accuracy of locating anatomical landmarks and identifying local 3D coordinate systems. As a result, kinematic measurements from biomechanical experiments can be applied to computer models.

Three flexion angles were examined at 30° increments. Smaller increments would have provided additional data points. However, the GRASS pulse (Gradient Refocused Acquisition in the Steady State) sequence was time consuming for each scan thereby limiting data collection. Moreover, the confining measurement volume (i.e. MR chamber) did not allow flexion to 90°.

Measurements were obtained under static non-physiological loading conditions with subjects supine in the MR chamber. Thus loading and the effects of muscular stabilisation were not considered which limits the interpretations of our findings. The orthosis used for accurate positioning ensured the knee did not move during data acquisition. The positioning device aided in alignment but it did not externally manipulate the knee into a position that would confound the results. Muhle et al. [92] reported that such devices when employed in a clinical setting have proven helpful in guiding joint motion while achieving sufficient image quality for kinematic MR studies.

TIBIOFEMORAL JOINT MOTION DERIVED FROM BONE PINS

The ACL deficient group vs. normal controls

Recording knee rotations and translations using instrumented spatial linkage systems and surface markers are not as accurate as bone pins secured directly into the skeleton. To date, the pin technique provides the most sensitive and accurate means to measure skeletal tibiofemoral joint motion. It may also provide a more sensitive measure of the differences between bracing conditions.

Although study I included a small number of subjects, intra-subject tibiofemoral rotations and translations showed a general trend across conditions, i.e. the shape and amplitudes of the skeletal marker based curves were similar. Patterns corresponded well with previous in-vivo tibiofemoral investigations although magnitudes differed [77,89,109,110]. The discrepancies across investigations are likely the result of differences in activity and to differences in the placement of the segmental anatomical axes. Lafortune [77] and McClay [89] employed anatomical coordinate systems based on a roentgen-stereophotogrammetric analysis whereas Reinschmidt [108,109,110] utilized the neutral standing trial.

The offset in the kinematic patterns between bracing may be due to the brace. But it 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. This created small deviations in alignment of the tibial and femoral anatomical coordinate systems. This is particular to *in-vivo* investigations using pins whereby differences in defining of the anatomical co-ordinate system may account for the variations in secondary rotations and translations [89]. 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.

Without brace, anterior tibial displacements ranged 2.2 to 8.8 mm (mean 4.4 mm) whereas with the brace, displacements were between 2.4 to 5.7 mm (mean 3.6 mm). Although it was not our intent to load the knee and create a giving way episode, one subject experienced tibial displacements of 8.8 mm. The negligible reductions in intra-subject and mean anterior displacements suggest no substantial mechanical effect with the DonJoy Legend brace. However, this may be due to the invasiveness of this protocol, that landings are performed onto a deficient limb, or that subjects jumped within their own comfort limits and did not adequately provoke the knee to give way.

In contrast, displacements for the normal controls in study III ranged 1.2 to 7.3 mm (mean 3.7 mm) for the jump and ranged 1.4 to 10.3 mm (mean 6.1 mm) for the cut. Our findings concur with Beynnon *et al*. [11] who reported the DonJoy Legend brace was incapable of reducing abnormal anterior tibial translations to within normal. They found that ACL deficient and normal knees underwent anterior translations of 4.6 mm (range 1.4 to 10.6 mm) and 1.3 mm (range 0.5 to 3.3 mm) respectively when changing between non-weightbearing and weightbearing postures. When supported with the DonJoy Legend brace, anterior translations of ACL deficient knee were on average 2.3 mm (3.1 mm), similar to the results in study III. It also confirms their earlier study that revealed ACL deficient knees on average experienced anterior translations of 3.4 mm (S.D. = 2.6 mm) compared to 0.8 mm (2.2 mm) for the contralateral normal knee [16]. This may explain why ACL deficient subjects report continued subluxation episodes or abnormal translations while using a functional brace. Internal loads generated from bodyweight and muscle contraction at foot-strike gait or when pivoting that the brace was not capable of reducing abnormal anterior translations to within limits of the normal joint [11].

The discrepancy among normal controls between studies can be attributed to differences in the loading conditions. Beynnon *et al*., [11] had subjects positioned supine in a laxity device [129] that simulated compressive knee loading. The normal controls in study III challenged the knee with muscle contraction and bodyweight when landing from the jump and during cutting. This is important because internal loads created by muscle contractions and compressive loading may influence displacements [11,139]. Moreover, anterior-posterior displacements between the tibial and femoral anatomical reference points may be influenced by the femoral "kink angle". As described in study V, when flexing the knee from terminal extension the femur "rocks" back as a result of the differing radii on the posterior femoral condyles [59,103]. Since the femoral anatomical coordinate system is located deep within the femoral notch, the femoral anatomical reference frame may shift posteriorly up to 8 mm and 2 mm on the medial and lateral condyles respectively, relative to the tibial anatomical frame [103]. This shift occurs within the femur and the femoral condyles themselves do not shift relative to the tibia. Because tibial translations were calculated relative to the femoral anatomical reference frame and that joint motion was described using the joint coordinate system (JCS) [47], a posterior femoral displacement would be described as an anterior tibial displacement. Therefore, a portion of anterior-posterior displacements along the floating axis may be attributed to this phenomenon.

Ground reaction force data when considered in combination with our brace study produced some unexpected results. Peak vertical force values for normal controls did not exceed 2 times bodyweight whereas the ACL deficient group experienced up to 3 times bodyweight [106]. The decline in peak force magnitudes diminished anterior displacements for the jump yet sizeable anterior displacements of 6.1 mm (range 1.4 to 10.3 mm) were observed for the cut. However, similar anterior translation magnitudes were observed for the non-braced and braced ACL deficient subjects and normal controls. Since the knee can experience peak vertical forces of up to eight times bodyweight during high dynamic activity, it was speculated that the level of joint loading might not have been adequate to provoke the large displacements necessary to identify kinematic differences.

EMG

Following ACL injury, compensatory muscle recruitment is required in maintaining joint stability [17,139]. The hamstrings and quadriceps work synergistically in protecting the joint [17]. It has been shown that bracing the ACL deficient knee evoked changes in EMG activity [2,17,95]. Perhaps the brace acts as a proprioceptive mechanism that influences afferent neural inputs to the central nervous system and mediates hamstring and quadriceps activity [95]. Mechanoreceptors about the joint, muscles and tendon and deep tissue convert peripheral information regarding joint motion, position, and muscle tension into neural impulse. Inputs are transmitted along afferent pathways to the central nervous system (CNS) to regulate neuromuscular control [80]. One might expect changes in muscle firing patterns, timing or reductions in amplitudes.

The transition between non-weightbearing and weightbearing was analysed separately to reduce the issue of variance [105]. Three time intervals were analysed: (A1) 250 ms preceding footstrike, (A2) 125 ms following footstrike and (A3) an additional 125 ms interval. Time intervals no less than 125 ms were selected in order to overcome the issue electromechanical delay. The 125 ms time interval (A2) was of primary interest, the period when the foot contacted the force plate and when peak Fy was attained.

Changes in muscle activity were observed as a result of bracing. The Friedman two-way ANOVA by rank for *k* related samples revealed a significant increase (21%) in rectus femoris activity at footstrike (A2) when the knee was braced. Moreover, significant decreases in both semitendinosus (17%) and biceps femoris (44%) activity were observed during periods A1 and A2 respectively. A reduction in hamstring activity with a concomitant increase in quadriceps activity would have the tendency to draw the tibia anteriorly during A2. Despite increased rectus femoris activity, there was no concomitant increase in anterior displacement of the tibia.

In study I, we reported no consistent reductions in anterior tibial translations as a result of bracing. We speculated the brace may allow for the generation of larger forces during strenuous activity and may not prevent abnormal tibial displacements When considered in combination with studies II and III, the results suggest increased afferent inputs from knee proprioceptors and the brace-skin-bone interface to the CNS evoke adaptive motor response and modify EMG activity [95]. In light of our findings with respect to changes in linear kinematics, it might be speculated that joint stability may result from proprioceptive feedback rather than the mechanical stabilising effect of the brace.

Because of the small sample size, we opted for the Friedman ANOVA by ranks. This test is a non-parametric alternative to one-way within-subject repeated measures ANOVA. Since each subject served as their own control, the Friedman ANOVA tests for differences between two or more dependent measures (or repeated testing). The point of using a matched test is to control for experimental variability between subjects, thus increasing the power of the test. Values in each matched set are first ranked then the ranks in each group are summed. If the sums are very different, the notion that differences are coincidences of random sampling can be rejected [114]. No inferences were drawn rather emphasis was to highlight significance to this particular group with respect to the one legged hop and the associated translations.

Inter-subject comparisons were not possible since the EMG data were not normalised to peak across the bracing conditions or to maximal voluntary isometric contractions. The major disadvantage lies in the measurement scales. This provides no information on the degree of muscular activation upon landing. With no reference to the subject's capacity, whether the reference level is 10% or 90% of the subject's maximum capacity is unknown. Knowledge of the proportion of a subject's muscle capacity required to perform a task may be important.

In examining changes in muscular-activation patterns among patient's with an old anterior cruciate ligament rupture, Gauffin *et al.* [44] found different muscular-activation pattern for the injured leg compared to the non-injured when performing a One-Legged Hop for distance at the moment of landing. Contrary to our methodology, subjects jumped from their deficient limb and landed onto the same leg. Quadriceps activity was lower for the ACL deficient group but no differences in the hamstring muscles were observed. Although subjects were well rehabilitated, there were minor deficits in muscle strength that may be a possible cause for differences in measurements. Differences may also be the result of the different techniques in performing the jump. By jumping from the deficient limb, a significant amount of pretension will be incurred in the musculature prior to landing. Conversely, with the modified one legged jump, the deficient limb would be more relaxed. Branch *et al.* [17,18] reported increased hamstring activity with a concomitant reduction in quadriceps activity during stance when performing side-step step cutting manoeuvres. This could have a protective effect on a deficient knee while affecting performance.

INTRA-CORTICAL PIN TECHNIQUE

After the first subject bent the pin, longer longitudinal incisions were made about the femoral insertion site and deep flexion and full extension movements were restricted. It was presumed impingement complications would be overcome. However the problem was not rectified as bending continued.

It is evident the Iliotibial band and the quadriceps tendon can generate force to bend the pin. The combination of knee flexion/extension and internal/external rotation of the tibia about the femur may account for the bend. Tibiofemoral joint motion has been described as spiral or helicoid during flexion and extension [101]. 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° [101]. 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 but it could account for bending the femoral pin.

A solution would be to use pins with a larger diameter and longer longitudinal incisions about the femoral insertion site. However, any impingement problem that may be encountered when using the larger diameter pin could cause injury to the surrounding tissue, rather than the pin deforming. A second solution would be to have pins with a shorter threaded section of about 10-15 mm. This would result in the threads being completely inserted into the cortice of the femur while exposing the more bending resistant area of the pin. Other means of avoiding the iliotibial band is to place the pins and markers on the medial side [77,81]. However the pins and target markers would interfere with the contralateral leg, particularly in subjects with close knee contact during walking [77]. To overcome this, Lafortune *et al.* [77] directed the target markers upward and anterior to the thigh using an extension rod to prevent the target markers from interfering with the contralateral leg during walking. To negate moments generated by the extension rod and target cluster, a second rod in the opposite direction served as a counter weight. However, the shock from foot-strike during strenuous activity introduced unacceptable vibrations to the extension rods as a result of the longer lever arms [74,78] Alternatively, Lundberg [87] suggested making a slit incision in the iliotibial tract to avoid restriction of flexion to avoid impingement problems [110].

Implications for pin research

Generalizations cannot be drawn as a result of the small sample size. Moreover, application of this methodology is limited, primarily due to its invasiveness. One cannot exclude the fact that the local anaesthesia, incisions into the muscle tissue and introduction of the pins might introduce tension thereby adversely affecting or inhibiting performance and electromyographic activity. However subjects reported they did not experience significant discomfort and all reported they moved their knees freely.

In combination with studies I and II, with respect to changes in linear kinematics (study I), it might be speculated that joint stability may result from proprioceptive feedback rather than the mechanical stabilising effect of the brace. When considering normative data in study III, the DonJoy Legend brace was incapable of reducing abnormal anterior tibial translations to within normal when changing between non-weightbearing and weightbearing postures. However the level of activity may not have been adequate to provoke the large displacements necessary to identify whether knee braces reduced pathological translations. And finally, since the DonJoy Legend brace was solely used in this study one may not be certain that the results can be applicable to other functional braces. Further research is required to verify these findings.

With the advancement of imaging technologies, surface marker solidification techniques and the development of algorithms to correct for movement artefacts, it is conceivable that the use of bone pins will eventually be eliminated. In fact, this technique may be one day used to validate other skeletally based kinematic measurement protocols. Until then, intra-cortical bone pins are the most accurate means of measuring tibiofemoral kinematics under dynamic conditions. Although there are inherent difficulties and risks involved with improvements in the insertion this technique can become safer and more reliable.

BONY CONTACT POINTS DERIVED FROM KINEMATIC MRI

Movement of the contact points between the medial and lateral compartments was asynchronous [143]. Their position varied with flexion. From 0° to 30°, the centroid of the contact area translated posteriorly, laterally and inferiorly on the tibial condyles along with a corresponding external tibial rotation. The predominant posterior displacement suggests femoral rolling. The posterior and inferior deflections may result from a combination of factors. The posterior femoral condyles have been described as having two circular arcs; a small anterior extension facet and a large posterior flexion facet [59,103]. The differing radii between the extension and flexion facets at their junction produce a “kink angle”. When flexing the knee from terminal extension, the femur “rocks” back from the extension facet onto the flexion facet. As a consequence, the medial tibial contact point reportedly shifted posteriorly, up to 8 mm [103]. This shift occurs within the femur and the femoral condyles themselves do not shift relative to the tibia. Since our results were obtained with the knee unloaded, posterior displacements of 12.5 mm and 13.8 mm on the lateral and medial condyle with the knee flexed 30° may not be an overestimation. Moreover, the contact point was drawn down along the sloping tibial plateau. Inferior displacements of up to 2.7 mm and 1.9 mm for the medial and lateral tibiofemoral have been previously reported during flexion [51].

Moving from 30° to 60°, the femur continued laterally but experienced a superior inflection. Only the lateral contact point displaced anteriorly, indicative that anteroposterior motion along medial femoral condyle was almost pure sliding. The posterior displacements and the smaller magnitudes seen medially during early flexion are in general agreement with previous studies although magnitudes differ [100].

It is likely that the differences in magnitudes are attributed to the methodology in deriving the location of the tibiofemoral contact points. Pinskerov *et al.* [103], Iwaki *et al.* [59] and Hill *et al.* [51] derived the contact points by identifying the “instant” geometric centre of the posterior femoral condyles (anterior and posterior flexion facets) and drawing a vertical line perpendicular to the tibial contact surface at each point in flexion. Anteroposterior motion was then expressed as a change in the position relative to the anatomical reference point. The drawback to their analysis was that is was restricted to sagittal plane movement. Conversely we identified the contact points as the centroid of the contact area. Secondly, our measurements were obtained under static non-physiological loading conditions. Subjects lied supine in the MR chamber and performed an open muscular chain rather than a closed muscular chain. Joint loading and timely muscle contractions may alter forces acting on the tibiofemoral plateau thereby creating new contact area profiles and pressure distributions.

Using the data collected from force plate, further studies can focus on deriving the location and orientation of the force vector in relation to the contact points. By combining the passive knee muscle moment arm data obtained from MRI [142], a comprehensive biomechanical model can be developed in order to calculate internal knee joint forces during the modified one legged jump and lateral side cut.

CONCLUSIONS

In using intra-cortical pins inserted into the tibia and femur of normal controls and patients with ACL deficiency, accurate descriptions of in-vivo tibiofemoral joint motion was achieved.

While the authors acknowledge that generalizations are not possible as a result of the small sample size, the techniques used in this investigation have provided a valuable insight into the three-dimensional kinematic behavior of the normal and ACL deficient knee joint during moderate to intense activity.

Non-customized (generic) functional knee braces brace are incapable of reducing abnormal anterior tibial translations to within normal when changing between non-weightbearing and weightbearing postures.

It may be suggested that joint stability may result from proprioceptive feedback rather than the mechanical stabilising effect of the brace. Moreover, we suggest an increase in afferent inputs from knee proprioceptors and brace-skin-bone interface modifies EMG activity.

The forces generated from biological tissue to permanently deform the bone pins were relatively small. During activity these forces are often exceeded. Therefore, in controlled experimental settings whereby bone pins will be utilized for deriving skeletal joint motions, careful attention is required for corrected pin orientation and insertion to reduce the risk of impingement and bending.

When performing unloaded open chain knee flexion movements, the position of the contact points varied with flexion. Moreover motion was not symmetrical between the medial and lateral knee compartments. The larger posterior displacements on the lateral condyle than medially suggests combined rotation of the tibia during flexion.

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REFERENCE LIST

[1]. Abdel-Aziz YI, Karara MH. Direct Linear Transformation from Comparator Coordinates into Object Space Coordinates in Close Range Photogrammetry. Proceedings A.S.P Symposium on Close Range Photogrammetry 1971;American Society of Photogrammetry:1-19

[2]. Acierno SP, D'Ambrosia C, Solomonow M, Baratta RV, et al. Electromyography and biomechanics of a dynamic knee brace for anterior cruciate ligament deficiency. Orthopedics 1995;18(11):1101-7

[3]. Ahmed AH, Burke D.L., Duncan N.A., Chan K.H. Ligament Tension in the Flexed Knee Combined Passive Anterior Translation and Axial Rotation. J Orthop Res 1992;10(6):854-67

[4]. Anderson K, Wojtys EM, Loubert PV, Miller RE. A biomechanical evaluation of taping and bracing in reducing knee joint translation and rotation. Am J Sports Med 1992;20(4):416-21

[5]. Andriacchi TP, Alexander EJ, Toney MK, Dyrby C, et al. A point cluster method for in vivo motion analysis: applied to a study of knee kinematics. J Biomech Eng 1998;120(6):743-49

[6]. Areblad M, Nigg BM, Ekstrand J, Olsson KO, et al. Three-dimensional measurement of rearfoot motion during running. J Biomech 1990;23(9):933-40

[7]. Asano T, Akagi M, Tanaka K, Tamura J, et al. In vivo three-dimensional knee kinematics using a biplanar image- matching technique. Clin Orthop Rel Res 2001;388:157-66

[8]. Baltzopoulos V. A videofluoroscopy method for optical distortion correction and measurement of knee-joint kinematics. Clin Biomech 1995;10(2):85-92

[9]. Banks SA, Hodge WA. Accurate measurement of three-dimensional knee replacement kinematics using single-plane fluoroscopy. IEEE Trans Biomed Eng 1996;43(6):638-49

[10]. Baratz ME, Fu FH, Mengato R. Meniscal tears: The effect of meniscectomy and of repair on intraarticular contact areas and stress in the human knee. A preliminary report. Am J Sports Med 1986;14(4):270-275

[11]. Beynnon BD, Fleming BC, Churchill D.L, Brown LE. The effect of anterior cruciate ligament-deficiency and functional bracing on translation of the tibia relative to the femur during non-weightbearing and weightbearing. Am J Sports Med 2003;31(1):in press

[12]. Beynnon BD, Fleming BC, Johnson RJ, Nichols CE, et al. Anterior cruciate ligament strain behavior during rehabilitation exercises in vivo. Am J Sports Med 1995;23(1):24-34

[13]. Beynnon BD, Johnson RJ, Fleming BC, Peura GD, et al. The effect of functional knee bracing on the anterior cruciate ligament in the weightbearing and nonweightbearing knee. Am J Sports Med 1997;25(3):353-59

[14]. Beynnon BD, Johnson RJ, Fleming BC, Renstrom PA, et al. The measurement of elongation of anterior cruciate-ligament grafts in vivo. J Bone Joint Surg Am 1994;76(4):520-531

[15]. Beynnon BD, Pope MH, Wertheimer CM, Johnson RJ, et al. The effect of functional knee-braces on strain on the anterior cruciate ligament in vivo. J Bone Joint Surg Am 1992;74(9):1298-312

[16]. Beynnon BD, Fleming BC, Labovitch R, Parsons B. Chronic anterior cruciate ligament deficiency is associated with increased anterior translation of the tibia during the transition from non-weightbearing to weightbearing. J Orthop Res 2002;20(2):332-37

[17]. Branch TP, Hunter R, Donath M. Dynamic EMG analysis of anterior cruciate deficient legs with and without bracing during cutting. Am J Sports Med 1989;17(1):35-41

[18]. Branch TP, Hunter RE. Functional analysis of anterior cruciate ligament braces. Clin Sports Med 1990;9(4):771-97

[19]. Brandsson S, Karlsson J, Eriksson BI, Karrholm J. Kinematics after tear in the anterior cruciate ligament: dynamic bilateral radiostereometric studies in 11 patients. Acta Orthop Scand 2001;72(4):372-78

[20]. Cappozzo A, Catani F, Leardini A, Benedetti MG, et al. Position and orientation in space of bones during movement: Experimental artefacts. Clin Biomech 1996;11:90-100

[21]. Cappozzo A. Three dimensional analysis of human walking: experimental methods and associated artifacts. Hum Mvmt Sci 1991;10(5):589-602

[22]. Cappozzo A, Catani F, Della Croce U, Leardini A. Position and orientation in space of bones during movement: anatomical frame definition and determination. Clin Biomech 1995;10:171-78

[23]. Cawley PW, France EP, Paulos LE. The current state of functional knee bracing research. A review of the literature. Am J Sports Med 1991;19(3):226-33

[24]. Chao EY, Laughman RK, Schneider E, Stauffer RN. Normative data of the knee joint motion and ground reaction forces in adult level walking. J Biomech 1983;16:219-33

[25]. Cheng CH, Yao NK, Liu HC, Lee KS. Influences of configuration changes of the patella on the knee extensor mechanism. Clin Biomech 1996;11:116-20

[26]. Cheng CK, Yao NK, Liu HC. Surgery simulation analysis of anterior advancement of the tibial tuberosity. Clin Biomech 1995;10:115-21

[27]. Cheng CK, Yao NK, Liu HC Weight bearing influence on knee joint bony contact movements: an in vivo video-fluoroscopy study. In: Proceedings of the National Science Council, Republic of China - Part B, Life Sciences. 20-25

[28]. Cheze L, Fregly BJ, Dimnet J. A solidification procedure to facilitate kinematic analyses based on video system data. J Biomech 1995;28(7):879-84

[29]. Ciccotti MG, Kerlan RK, Perry J, Pink M. An electromyographic analysis of the knee during functional activities. II. The anterior cruciate ligament-deficient and -reconstructed profiles. Am J Sports Med 1994;22(5):651-58

[30]. Cook FF, Tibone JE, Redfern FC. A dynamic analysis of a functional brace for anterior cruciate ligament insufficiency. Am J Sports Med 1989;17(4):519-24

[31]. Davis MP, Öunpuu S, Tyburski D, Galante JO. A gait analysis data collection and reduction technique. Hum Mvmt Sci 1991;10:575-87

[32]. DeLuzio KJ, Wyss UP, Li J, Costigan PA. A procedure to validate three-dimensional motion assessment systems. J Biomech 1993;26(6):753-59

[33]. Deutsch A, Pressman BD, Shellcock FG, Mink JH. Kinematic magnetic resonance imaging of the joints: Techniques and clinical applications. Magn Reson Q 1991;7:104-35

[34]. DeVita P, Hunter PB, Skelly WA. Effects of a functional knee brace on the biomechanics of running. Med Sci Sports Exerc 1992;24(7):797-806

[35]. Draganich LF, Jaeger RJ, Kralj AR. Coactivation of the hamstrings and quadriceps during extension of the knee. J Bone Joint Surg Am 1989;71(7):1075-81

[36]. Eastlack ME, Axe MJ, Snyder-Mackler L. Laxity, instability, and functional outcome after ACL injury: copers versus noncopers. Med Sci Sports Exerc 1999;31(2):210-215

[37]. Elmqvist LG, Lorentzon R, Johansson C, Fugl-Meyer AR. Does a torn anterior cruciate ligament lead to change in the central nervous drive of the knee extensors? Eur J Appl Physiol 1988;58(1-2):203-7

[38]. Fleming BC, Beynnon BD, Nichols CE, Johnson RJ, et al. An in vivo comparison of anterior tibial translation and strain in the anteromedial band of the anterior cruciate ligament. J Biomech 1993;26(1):51-58

[39]. Frank CB, Shrive NG. Biomaterials: Ligament. (In: Nigg BM, Herzog W., eds. Biomechanics of Musculo-skeletal system.) New York: John Wiley & Sons, 1994; 106-32.

[40]. Fukubayashi T, Kurosawa H. The contact area and pressure distribution pattern of the knee. A study of normal and osteoarthrotic knee joints. Acta Orthop Scand 1980;51(6):871-79

[41]. Fuller J, Liu L-J, Murphy MC, Mann RW. A comparison of lower-extremity skeletal kinematics measured using skin- and pin-mounted markers. Hum Mvmt Sci 1997;16(2-3):219-42

[42]. Gauffin H, Petterson G., Tropp H. Kinematic analysis of one-leg long hopping in patients with an old rupture of the anterior cruciate ligament. Clin Biomech 1990;11:73-77

[43]. Gauffin H, Pettersson G, Tegner Y, Tropp H. Function testing in patients with old rupture of the anterior cruciate ligament. Int J Sports Med 1990;11(1):73-77

[44]. Gauffin H, Tropp H. Altered movement and muscular-activation patterns during the one legged jump in patients with an old anterior cruciate ligament rupture. Am J Sports Med 1992;20(2):182-92

[45]. Gillquist J. Knee ligaments and proprioception. Acta Orthop Scand 1996;67(6):533-35

[46]. Granberry WM, Noble PC, Woods GW. Evaluation of an electrogoniometric instrument for measurement of laxity of the knee. [see comments.]. Journal of Bone & Joint Surgery 1990;72(9):1316-22

[47]. Grood EW, Suntay WJ. A Joint Coordinate System for the Clinical Description of Three Dimensional Motions: Applications to the Knee. J Biomech Eng 1983;105:136-44

[48]. Gryzlo SM, Patek RM, Pink M, Perry J. Electromyographic analysis of knee rehabilitation exercises. J Orthop Sports Phys Ther 1994;20(1):36-43

[49]. Hale JE, Brown TD. Contact stress gradient detection limits of Pressensor film. J Biomech Eng 1992;114:352-57

[50]. Harris GF, Wertsch JJ. Procedures for gait analysis. Arch Phys Med Rehabil 1994;75(2):216-25

[51]. Hill PF, Vedi V, Williams A, Iwaki H, et al. Tibiofemoral movement 2: The loaded and unloaded living knee studied by MRI. J Bone Joint Surg Br 2000;82(8):1196-98

[52]. Hille E, Schulitz KP, Henrichs C, SchneiderT. Pressure and contract-surface measurements within the femoropatellar joint and their variations following lateral release. Arch Orthop Trauma Surg 1985;104(5):275-82

[53]. Hirokawa S, Solomonow M, Lu Y, Lou ZP, et al. Anterior-posterior and rotational displacement of the tibia elicited by quadriceps contraction. Am J Sports Med 1992;20(3):299-306

[54]. Hofmann AA, Wyatt RW, Bourne MH, Daniels AU. Knee stability in orthotic knee braces. Am J Sports Med 1984;12(5):371-74

[55]. Howell DC. Statistical Methods for Psychology. 3rd Belmont: Duxbury Press, 1992

[56]. Ihn JC, Kim SJ, Park IH. In vitro study of contact area and pressure distribution in the human knee after partial and total meniscectomy. Int Orthop 1993;17(4):214-18

[57]. Inaba H, Arai M. A method for measuring contact pressures instantaneously in articular joints. J Biomech 1989;22(11/12):1293-96

[58]. Ishii Y, Terajima K, Terashima S, Koga Y. Three-dimensional kinematics of the human knee with intracortical pin fixation. Clin Orthop Rel Res 1997;343:144-50

[59]. Iwaki H, Pinskerova V, Freeman MA. Tibiofemoral movement 1: The shapes and relative movements of the femur and tibia in the unloaded cadaver knee. J Bone Joint Surg Br 2000;82(8):1189-95

[60]. Johansson H, Sjolander P, Sojka P. A sensory role for the cruciate ligaments. [Review]. Clin Orthop Rel Res 1991;(268):161-78

[61]. Johansson H, Sjolander P, Sojka P. Receptors in the knee joint ligaments and their role in the biomechanics of the joint. [Review]. Crit Rev Biomed Eng 1991;18(5):341-68

[62]. Jonsson H, Karrholm J. Brace effects on the unstable knee in 21 cases. A roentgen stereophotogrammetric comparison of three designs. Acta Orthop Scand 1990;61(4):313-18

[63]. Jonsson H, Karrholm J. Three-dimensional knee joint movements during a step-up: evaluation after anterior cruciate ligament rupture. J Orthop Res 1994;12(6):769-79

[64]. Jonsson H, Karrholm J, Elmqvist LG. Kinematics of active knee extension after tear of the anterior cruciate ligament. Am J Sports Med 1989;17(6):796-802

[65]. Kadaba MP, Ramakrishnan HK, Wootten ME. Measurement of lower extremity kinematics during level walking. J Orthop Res 1990;8(3):383-92

[66]. Karlsson D., Lundberg A. In vivo measurement of the shoulder rhythm using external fixation markers. In: 3rd International Symposium on 3-D Analysis of Human Movement. Hasselbacken Conference Centre, Stockholm 69-72

[67]. Karlsson D, Tranberg R. On skin movement artefact-resonant frequencies of skin markers attached to the leg. Hum Mvmt Sci 1999;18(5):627-35

[68]. Karrholm J, Elmqvist LG, Selvik G, Hansson LI. Chronic anterolateral instability of the knee. A roentgen stereophotogrammetric evaluation. Am J Sports Med 1989;17(4):555-63

[69]. Karrholm J, Selvik G, Elmqvist LG, Hansson LI, et al. Three-dimensional instability of the anterior cruciate deficient knee. Journal of Bone & Joint Surgery - British Volume 1988;70(5):777-83

[70]. Kaufman KR, Kovacevic N, Irby SE, Colwell CW. Instrumented implant for measuring tibiofemoral forces. J Biomech 1996;29(5):667-71

[71]. Kennedy PW, Wright DL, Smith GA. Comparison of film and video techniques for three-dimensional DLT repredictions. Int J Sports Biomech 1989;5:457-60

[72]. Kettelkamp DB, Jacobs AW. Tibiofemoral contact area--determination and implications. J Bone Joint Surg Am 1972;54(2):349-56

[73]. Klien PJ, DeHaven JJ. Accuracy of three dimensional linear and angular estimates obtained with the Ariel Performance System. Arch Phys Med Rehabil 1995;76(2):183-89

[74]. Koh TJ, Grabiner MD, De SRJ. In vivo tracking of the human patella. J Biomech 1992;25(6):637-43

[75]. Krauspe R, Schmidt M, Schaible HG. Sensory innervation of the anterior cruciate ligament. An electrophysiological study of the response properties of single identified mechanoreceptors in the cat. J Bone Joint Surg Am 1992;74(3):390-397

[76]. Lafortune MA. The Use of Intra-cortical Pins to Measure the Motion of the Knee Joint During Walking. Doctoral Thesis, The Pennsylvania State University, 1984.

[77]. Lafortune MA, Cavanagh PR, Sommer HJ, Kalenak A. Three-dimensional kinematics of the human knee during walking. J Biomech 1992;25(4):347-57

[78]. Lafortune MA, Cavanagh PR, Sommer HJ, Kalenak A. Foot inversion-eversion and knee kinematics during walking. J Orthop Res 1994;12(3):412-20

[79]. Lenox JB, Cuzzi JR. Accurately Characterizing a Measured Change in Configuration. ASME paper 1978;No.78-DETY-50:

[80]. Lepart SM, Riemann BL, Fu FH. Introduction to the sensorimotor system. (In: Lephart SM, Fu FH, eds. Proprioception and Neuromuscular Control in Joint Stability.) Champaign IL: Human Kinetics Publishers, 2000; xvii-xviv.

[81]. Levens AS, Inman VT, Blosser JA. Transverse Rotation of the Segments of the Lower Extremity in Locomotion. J Bone Joint Surg 1948;30A:859-72

[82]. Liu SH, Lunsford T, Gude S, Vangsness CT, Jr. Comparison of functional knee braces for control of anterior tibial displacement. Clinical Orthopaedics & Related Research 1994;(303):203-10

[83]. Lucchetti L, Cappozzo A, Cappello A, Della CU. Skin movement artefact assessment and compensation in the estimation of knee-joint kinematics. J Biomech 1998;31(11):977-84

[84]. Lundberg A, Bylund C, Selvik G, Winson IG. Accuracy of Roentgen stereophotogrammetric analysis in joint kinematics. Eur J Exp Musculoskel Res 2002;7:213-16

[85]. Lundberg A. Kinematics of the ankle and foot: in-vivo roentgen stereophotogrammetry. Acta Orthop Scand 1989;60:291-96

[86]. Lundberg A, Winson IG, Nemeth G, Josephson A. In Vitro Assessment of the Accuracy of Opto-electric Joint Motion Analysis. Eur J Exp Musculoskel Res 1992;1:217-19

[87]. Lundberg A. On the use of bone and skin markers in kinematics research. Hum Mvmt Sci 1996;15(3):411-22

[88]. Marans HJ, Jackson RW, Glossop ND, Young C. Anterior cruciate ligament insufficiency: a dynamic three-dimensional motion analysis. Am J Sports Med 1989;17(3):325-32

[89]. McClay IS. A Comparisson of Tibiofemoral and Patellofemoral Joint Motion in Runners With and Without Patellofemoral Pain. Doctoral Thesis, The Pennsylvania State University, 1990.

[90]. More RC, Karras BT, Neiman R, Fritschy D, et al. Hamstrings--an anterior cruciate ligament protagonist. An in vitro study. Am J Sports Med 1993;21(2):231-37

[91]. Moseley L, Smith R, Hunt A, Gant R. Three-dimensional kinematics of the rear foot during the stance phase of walking in young adult males. Clin Biomech 1996;11:39-45

[92]. Muhle C, Brossmann J, HellerM. Kinematic MRI of the knee using a specially designed positioning device. J Comput Assist Tomogr 1996;20(4):522-25

[93]. Murphy MC. Geometry and Kinematics of the Normal Human Knee. Doctoral Thesis, The Massachusetts Institute of Technology, 1990.

[94]. Nakagawa S, Kadoya Y, Todo S, Kobayashi A, et al. Tibiofemoral movement 3: Full flexion in the living knee studied by MRI. J Bone Joint Surg Br 2000;82(8):1199-200

[95]. Németh G, Lamontagne M, Tho KS, Eriksson E. Electromyographic activity in expert downhill skiers using functional knee braces after anterior cruciate ligament injuries. Am J Sports Med 1997;25(5):635-41

[96]. Nigg BM, Cole GK. Optical Methods. (In: Nigg BM, Herzog W, eds. Biomechanics of the Musculo-Skeletal System.) New York: John Wiley & Sons, 1994; 254-86.

[97]. Nigg BM, Cole GK, Nachbauer W. Effects of arch height of the foot on angular motion of the lower extremities in running. J Biomech 1993;26(8):909-16

[98]. Niitsu M, Ikeda K, Fukubayashi T, Anno I, et al. Knee extension and flexion: MR delineation of normal and torn anterior cruciate ligaments. J Comput Assist Tomogr 1996;20(2):322-27

[99]. Niitsu M, Ikeda K, Itai Y. Slightly flexed knee position within a standard knee coil: MR delineation of the anterior cruciate ligament. Eur J Radiol 1998;8(1):113-15

[100]. Nisell R, Nemeth G, Ohlsen H. Joint forces in extension of the knee. Analysis of a mechanical model. Acta Orthop Scand 1986;57(1):41-46

[101]. Nordin M, Frankel VH. Basic Biomechanics of the Musculoskeletal System. 2nd Philadelphia: Lea & Febiger, 1989

[102]. O'Malley M, de Paor D. Kinematics of Human walking gait using digital image processing. Med Biol Eng Comput 1993;31(4):392-98

[103]. Pinskerova V, Iwaki H, Freeman MA. The shapes and relative movements of the femur and tibia in the unloaded cadaveric knee: A study using MRI as an anatomic tool. (In: Insall JN, Scott R, eds. Surgery of the Knee.) 3rd Ed. Philadelphia: W.B. Saunders, 2001; 255-83.

[104]. Ramakrishnan HK, Kadaba MP. On the estimation of joint kinematics during gait. J Biomech 1991;10:969-77

[105]. Ramsey DK, Wretenberg PF, Lamontagne M, Németh G. Electromyographic and biomechanic analysis of anterior cruciate ligament deficiency and functional knee bracing. 18 2003;1:28-34

[106]. Ramsey DK, Lamontagne M, Wretenberg PF, Valentin A, et al. Assessment of functional knee bracing: an in vivo three-dimensional kinematic analysis of the anterior cruciate deficient knee. Clin Biomech 2001;16(1):61-70

[107]. Ramsey DK, Wretenberg PF. Biomechanics of the knee: methodological considerations in the in vivo kinematic analysis of the tibiofemoral and patellofemoral joint. Clin Biomech 1999;14(9):595-611

[108]. Reinschmidt C. Three Dimensional Tibiocalcaneal and Tibiofemoral Kinematics During Human Locomotion - Measured with External and Bone Markers. Doctoral Thesis, The University of Calgary, Calgary, Canada, 1996.

[109]. Reinschmidt C, van den Bogert AJ, Nigg BM, Lundberg A, et al. Effect of skin movement on the analysis of skeletal knee joint motion during running. J Biomech 1997;30(7):729-32

[110]. Reinschmidt C, van den Bogert AJ, Lundberg A, Nigg BM, et al. Tibiofemoral and tibiocalcaneal motion during walking: external vs. skeletal markers. Gait Posture 1997;6(2):98-109

[111]. Renström P, Arms SW, Stanwyck TS, Johnson RJ, et al. Strain within the anterior cruciate ligament during hamstring and quadriceps activity. Am J Sports Med 1986;14(1):83-87

[112]. Ronsky JL, Herzog W, Brown TD, Pedersen DR, et al. In vivo quantification of the cat patellofemoral joint contact stresses and areas. J Biomech 1995;28(8):977-83

[113]. Rudert MM, Baratz ME, Rehak DC, Fu FH. Loading characteristics of pressensor. Transactions of the 34th Annual Meeting Orthop Res Soc 1988;1:226-27

[114]. Seigal S., Castellan NJ. Nonparametric Statistics for the Behavioral Sciences International Edition. 2nd Ed New York: McGraw-Hill Book Co., 1988

[115]. Selvik G. Roentgen stereophotogrammetry. A method for the study of the kinematics of the skeletal system. [Review] [42 refs]. Acta Orthop Scand 1989;Supplementum. 232:1-51

[116]. Selvik G, Alberius P, Aronson AS. A roentgen stereophotogrammetric system. Construction, calibration and technical accuracy. Acta Radiologica: Diagnosis 1983;24(4):343-52

[117]. Sheehan FT, Drace JE. Quantitative MR measures of three-dimensional patellar kinematics as a research and diagnostic tool. Med Sci Sports Exerc 1999;31(10):1399-405

[118]. Sheehan FT, Zajac FE, Drace JE. Using cine phase contrast magnetic resonance imaging to non-invasively study in vivo knee dynamics. J Biomech 1997;31(1):21-26

[119]. Sheehan FT, Zajac FE, Drace JE. In vivo tracking of the human patella using cine phase contrast magnetic resonance imaging. J Biomech Eng 1999;121(6):650-656

[120]. Smith BA, Livesay GA, Woo SL. Biology and biomechanics of the anterior cruciate ligament. Clin Sports Med 1993;12(4):637-70

[121]. Söderkvist I, Wedin PÅ. Determining the movements of the skeleton using well-configured markers. J Biomech 1993;26:1473-77

[122]. Spoor C, Veldpaus F. Rigid body motion calculated from spatial coordinates of markers. J Biomech 1980;21:45-54

[123]. Stiehl JB, Komistek R, Dennis DA. A novel approach to knee kinematics. Am J Orthop 2001;30(4):287-93

[124]. Stiehl JB, Komistek RD, Dennis DA, Paxson RD, et al. Fluoroscopic analysis of kinematics after posterior-cruciate-retaining knee arthroplasty. J Bone Joint Surg Br 1995;77(6):884-89

[125]. Takeda Y, Xerogeanes JW, Livesay GA, Fu FH, et al. Biomechanical function of the human anterior cruciate ligament. Arthroscopy 1994;10(2):140-147

[126]. Tashman S, Dupré K, Goitz H, Lock T, Kolowich P, Flynn M A digital radiographic system for determining 3D joint kinematics during movement. In: Proceedings of the 19th Annual Conference of the American Society of Biomechanics. The University of Stanford, Stanford 249-50

[127]. Tegner Y, Lysholm J. Rating systems in the evaluation of knee ligament injuries. Clin Orthop Rel Res 1985;198:43-49

[128]. Tibone JE, Antich TJ, Fanton GS, Moynes DR, et al. Functional analysis of anterior cruciate ligament instability. Am J Sports Med 1986;14(4):276-84

[129]. Un BS, Beynnon BD, Churchill DL, Haugh LD, et al. A new device to measure knee laxity during weightbearing and non-weightbearing conditions. J Orthop Res 2001;19(6):1185-91

[130]. Vailas JC, Pink M. Biomechanical effects of functional knee bracing. Practical implications. Sports Med 1993;15(3):210-218

[131]. Vailas JC, Tibone J.E., Perry J., Pink M. The Effects of Functional Knee Bracing on the Dynamic Biomechanics of the Cross Cut Manoeuver. J Sports Traumatol Rel Res 1990;12(4):197-206

[132]. van Dijk R, Huiskes R, Selvik G. Roentgen stereophotogrammetric methods for the evaluation of the three dimensional kinematic behaviour and cruciate ligament length patterns of the human knee joint. J Biomech 1979;12(9):727-31

[133]. Vergis A, Gillquist J. Sagittal plane translation of the knee during stair walking in healthy volunteers measured by an electrogoniometer chain. Scand J Med Sci Sports 1995;5(6):353-57

[134]. Vergis A, Gillquist J. Sagittal plane translation of the knee during stair walking. Comparison of healthy and anterior cruciate ligament--deficient subjects. Am J Sports Med 1998;26(6):841-46

[135]. Vergis A, Hindriks M, Gillquist J. Sagittal plane translations of the knee in anterior cruciate deficient subjects and controls. Med Sci Sports Exerc 1997;29(12):1561-66

[136]. Wallace WA, Johnson F. Detection and correction of geometrical distortion in X-ray fluoroscopic images. J Biomech 1981;14:123-25

[137]. Winter DA. Biomechanics and Motor Control of Human Movement. 2nd Ed. New York: Wiley-Interscience Publication, 1990

[138]. Winter DA, Yack H.J. EMG profiles during normal human walking: stride-to stride and inter-subject variability. Electroencephalogr Clin Neurophysiol 1987;67(5):402-11

[139]. Wojtys EM, Hutson H.J. Neuromuscular Performance in Normal and Anterior Cruciate Ligamnet Deficient Lower Extremities. Am J Sports Med 1994;22(1):89-104

[140]. Wojtys EM, Kothari SU, Huston LJ. Anterior cruciate ligament functional brace use in sports. Am J Sports Med 1996;24(4):539-46

[141]. Wojtys EM, Loubert PV, Samson SY, Viviano DM. Use of a knee-brace for control of tibial translation and rotation. A comparison, in cadavera, of available models. [see comments]. Journal of Bone & Joint Surgery - American Volume 1990;72(9):1323-29

[142]. Wretenberg P, Németh G, Lamontagne M, Lundin B. Passive knee muscle moment arms measured in-vivo with MRI. Clin Biomech 1996;11:439-46

[143]. Wretenberg P, Ramsey DK, Nemeth G. Tibiofemoral contact points relative to flexion angle measured with MRI. Clin.Biomech 2002;17(6):477-85

[144]. Wu JZ, Herzog W, Epstein M. Effects of inserting a pressensor film into articular joints on the actual contact mechanics. J Biomech Eng 1998;120(5):655-59

[145]. Yang JF, Winter DA. Electromyographic amplitude normalization methods: improving their sensitivity as diagnostic tools in gait analysis. Arch Phys Med Rehabil 1984;65(9):517-21