SKELETAL KINEMATICS OF THE ANTERIOR CRUCIATE LIGAMENT DEFICIENT KNEE WITH AND WITHOUT FUNCTIONAL BRACES

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Steinmann traction pins were implanted into the femur and tibia of six subjects having a partial or complete anterior cruciate ligament (ACL) rupture. Patients jumped for maximal horizontal distance and landed onto their deficient limb with the knee braced and unbraced. Tibiofemoral rotations and translations showed a general trend across subjects, i.e. skeletally based curves were similar in shape and amplitude. The tibia displaced anteriorly from footstrike to about peak vertical force onset (Fy). Thereafter the tibia moved posteriorly during flexion. Intra-subject kinematics was very repeatable but differences in anterior tibial translations were small between the brace conditions. This may be due to the invasiveness of this protocol, that landings were onto a deficient limb, or subjects jumped within their own comfort limits which did not maximally stress the ACL. Inter-subject differences were typically much larger.

KEY WORDS: knee, knee kinematics, brace, ACL, tibiofemoral, intra-cortical

INTRODUCTION: Functional knee braces are designed to stabilise anterior cruciate ligament (ACL) deficient knees by reducing pathological translations and rotations. Yet little research has examined the effects of these braces on three-dimensional osteokinematics and arthrokinetics during high dynamic activity. Braces are effective in reducing anterior translations when subjected to static or low anterior shear forces, but fail in situations where high loads are encountered or when the load is applied in an unpredictable manner . Recent investigations have employed target markers affixed to intra-cortical pins implanted into the tibia and femur to describe skeletal tibiofemoral joint motion . However, activities were restricted to walking or light running. Since braces are designed for athletic activity, they should be evaluated under such conditions. The purpose of this investigation was to determine whether application of a functional brace reduced rotational and linear tibial displacements during the performance of a One Legged Jump (OLJ).

METHODS: Six young normal healthy males diagnosed with partial or complete ACL rupture and having no prior surgical treatment were selected for this investigation. The Ethics Committee of the Karolinska Hospital approved the surgery and experiment. Steinmann traction pins were surgically implanted posterio-laterally into the femur and tibia with the knee flexed to 45°. No flexion-extension impairments resulted from impingements between the iliotibial band and the femoral pin or the brace/pin interface. Stereophotogrammetric radiographs (RSA) were taken with target markers affixed to the pins to identify the femoral and tibial anatomical reference points. The femoral anatomical 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 anatomical reference point. Kinematics were recorded with the MacReflex motion analysis system sampling at 120Hz within a 0.25m^3 measurement area (approximately 0.45 m off the floor). A Kistler force plate was synchronised to collect simultaneous ground reaction forces at 960 Hz. 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 0.25m^3 . To sufficiently stress the ACL, patients jumped for maximal horizontal distance. Subjects pushed off with their sound limb and landed onto the force platform with the contralateral deficient limb. The longest measurement was marked on the floor to determine the proper take-off distance to the force platform. Subjects were then randomly assigned to start with either the braced or non-braced condition. The brace (DonJoy Legend) was applied by the researcher according to the specifications prescribed by the manufacturer. Target marker orientations were recorded during a neutral standing trial to define the tibial and femoral anatomical co-ordinate system. The anatomical coordinate systems were arbitrarily defined to align with the global co-ordinate system during neutral standing. Five measurement trials and two neutral standing trials were recorded for each brace condition. Skeletal tibiofemoral joint kinematics were computed employing coordinate transformation matrices . Angular and linear motion was described as movement between the tibial anatomical reference frame relative to the femoral anatomical reference frame. Joint motion was referenced to the joint coordinate system (Grood & Suntay 1983). Cardan angles were employed to describe sequence of rotations and were computed about –y, x, z axes .

RESULTS & DISCUSSION: No subjects experienced significant discomfort and all reported they could move their knees freely. Data are presented for only four subjects. One subject was excluded due to the femoral pin bending during flexion. The second subject was excluded due to significant marker dropout in the kinematic data. Each subject served as their own control with analysis focusing on differences in magnitudes and changes in the shape of the curves between bracing conditions. Averages were derived for each subject during non-braced and braced testing. All force data were associated with the coincident kinematic frame number.

Kinetics. Intra-subject peak vertical force and peak posterior shear force was generally consistent between unsupported and braced conditions indicating that jumps onto the force platform were similar (Table 1). However, magnitudes varied across subjects. The differences between skeletal tibiofemoral kinematics across bracing conditions cannot be attributed to differences in jumping, but rather to the brace itself. Although the data recording system failed to store ground reaction force data for subject 6, angular data were used to determine whether jumping styles were similar between conditions.

Table 1 Means of Peak Vertical and Peak Posterior Ground Reaction Forces Normalised to Body Mass and Mass of the Brace Across Subjects and Conditions

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

Kinematics. As seen in Figure 1, an offset was evident between the unbraced and braced trials. This may be the result of the brace but is more likely the result of the different standing reference trials used for both test conditions. This created small deviations in alignment of the tibial and femoral anatomical co-ordinate systems. Therefore, differences in movement patterns were reported rather than the absolute positions, i.e. the range from touchdown to maximum flexion instead of the (absolute) maximum flexion value. All subjects demonstrated fairly similar flexion patterns although flexion ranges of motion varied (Table 2). With respect to the origins of the anatomical co-ordinate systems, anteroposterior curves were similar in shape between bracing conditions and fairly similar across subjects. The tibia exhibited a rapid anterior displacement with respect to the femur from footstrike to approximately peak

Flexion-extension **Anterior-posterior translations**

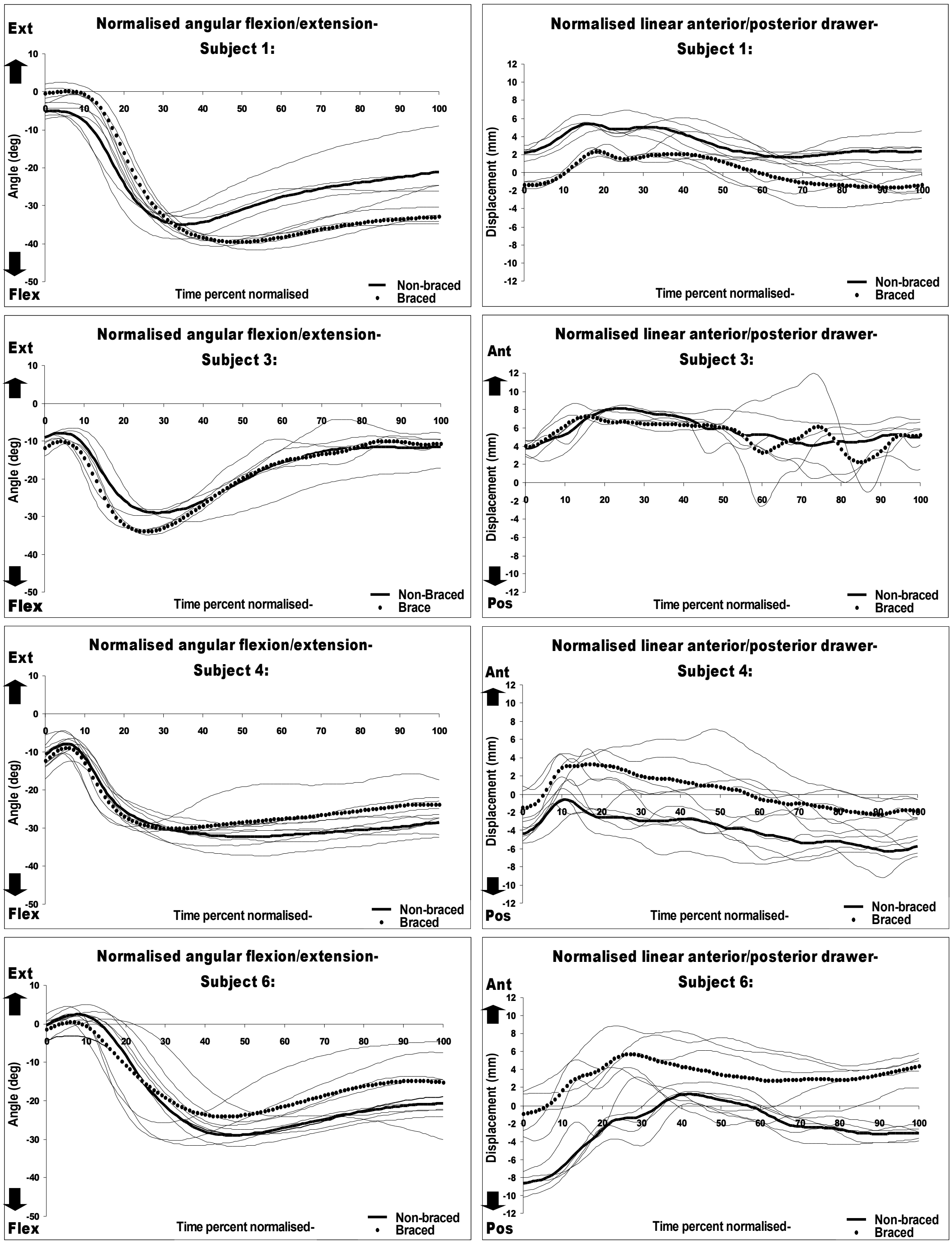

Figure 1 - **Flexion-extension and anterior-posterior patterns of tibiofemoral joint motion derived from skeletal (femur, tibia) markers. The averages of the trials are displayed in bold. The bold solid line represent the unbraced kinematics, the bold dashed line represent braced kinematics**.

Table 2 Means of Flexion and Anterior Translation Ranges of Motion

		Flexion		Anterior Draw	
Subject	Trials	Unbraced (degrees)	Braced (degrees)	Unbraced (mm)	Braced (mm)
1	n = 5	-29.9	-39.9	3.0	2.7
3	n = 3	-21.1	-23.7	3.5	2.4
4	n = 5	-24.2	-21.3	2.2	3.5
6	n = 5	-31.5	-24.6	8.8	5.7

i) A negative value indicates that flexion of the TFJ took place.
ii) A negative value indicates the tibia remained in a posterior position with respect to the femur even though it had moved in its most anteriorly located position.

vertical force Fy. Thereafter, the tibia was drawn posteriorly during flexion. However, differences in magnitudes between unbraced and braced patterns were small (Table 2). This may be due to the invasiveness of this protocol, that landings were onto a deficient limb, or subjects jumped within their own comfort limits which did not maximally stress the ACL. Generally, intra-subject knee kinematics were very repeatable. Tibiofemoral rotations and translations show a general trend across conditions, i.e. the shape and amplitudes of the skeletal marker-based curves were similar. As expected, inter-subject differences were typically much larger. Differences mainly consisted in amplitudes, orientation and position at footstrike. Additionally, patterns corresponded well with previous in-vivo tibiofemoral investigations although magnitudes differed. The discrepancies across investigations are likely the result of differences in locomotor activity and differences in the placement of the segmental anatomical axes.

CONCLUSION: The negligible reductions in anterior tibial draw indicates that the brace did not reduce translations during dynamic activity. From the lack of supportive evidence for bracing, a perceived improvement in performance may be the result of a proprioceptive feedback rather than the stabilising effect of the brace. The patient's subjective approval for the brace may allow for the generation of larger forces during strenuous activity but not prevent abnormal tibial displacements. The increase in forces acting at the knee are thought to accelerate the degenerative joint disease as seen in ACL deficient knees. Therefore, athletes who wear the brace during strenuous physical activity are at greater risk of generating increased forces and theoretically increase the risk of joint damage.

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