Effects of knee brace and AFO on knee stability: A case study
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INTRODUCTION:
One of the most common knee joint injuries in sport is the partial or total disruption of the anterior cruciate ligament (ACL)\(^1\). ACL-deficient knees can be unstable, give way unexpectedly and create significant problems during locomotion. Knee instability is currently treated with reconstructive surgery, or more conservatively, with therapy and the use of a functional knee orthosis (KO) to stabilize the knee. A KO is "designed to facilitate normal kinematics of the tibiofemoral joint while limiting abnormal displacement and loading."\(^2\)

Without the ACL to restrain anterior translation of the tibia, a reduction in the anterior force generated by quadriceps muscles may be advantageous.\(^3\) In the treatment of neurological quadriceps weakness, an ankle foot orthosis (AFO) is sometimes used.\(^4\) The AFO reduces the quadriceps activation necessary by creating a posterior force on the tibia, resulting in an applied extension moment at the knee. It was hypothesized that the use of an AFO to reduce quadriceps activation during gait would be beneficial to patients with knee instability due to ACL deficiency. A clinical trial of an ACL-deficient patient with a severely unstable knee using an AFO for stability was carried out at the Sport Medicine Center at the University of Calgary, Alberta, Canada. The patient reported that the AFO provided greater stability than the KO.

The purpose of this case study was to evaluate differences in this patient’s gait kinematics when using a KO or an AFO to stabilize the knee, and to compare them to normal gait. It was hypothesized that the difference between the AFO and the KO reported by the patient would be apparent in the kinematics and that the AFO would allow more normal gait kinematics. One particular aspect of the kinematics at the knee is the axis of rotation, which can be evaluated by calculating the finite helical axis (FHA).\(^5\) This method describes 3-D motion in terms of a rotation around a translation along one axis, the FHA. The pattern of change in the FHA over the gait cycle had been shown to be dependent on the motion\(^6\) but has never been used to evaluate the effects of bracing on gait. A sub-hypothesis of this study is that changes in knee motion due to the different bracing conditions would be evident in the path of the FHA.

METHODS:
One male participant with severe bilateral knee instability due to ACL deficiency and co-occurrent osteoarthritis was evaluated. Gait analysis was performed on the braced (left) leg while the subject, who required a cane to ambulate, walked in both the AFO and the KO. The AFO (Colman Prosthetics and Orthotics, Calgary, AB) was set at 25° plantar flexion, while the KO was an ACL brace for combined instabilities (Townsend Brace, Bakersfield, CA). Six volunteers (3 male, 3 female) with no history of neurological or musculo-skeletal dysfunction were also evaluated walking without assistive devices at 1.14±0.02 m/s. This established a control group to compare braced conditions against. The University of Calgary Conjoint Ethics Health Research Committee approved all human subject measurements.

One standard reflective marker set was used for the normal subjects and was altered for the braced subject due to the constraints of brace location. A fourth marker was added to each leg segment in the braced subject, to improve the reliability from the altered marker set. Video-based (Motion Analysis, Santa Rosa CA) movement data were collected (120 Hz), at a self-selected walking speed (0.61±0.03 m/s) for the braced conditions, and at a slow speed (1.14±0.02 m/s) for the normals. A trial of the subject standing aligned with the lab coordinate system was taken for each condition, and was used as a reference for the kinematic trials. Forceplate (Kistler Instrument Corp. Amherst, NY) data were collected at 1200 Hz.

Video data were smoothed at 4 Hz for accurate calculation of the FHA parameters, and normalized for one stance phase: from touch down to take off. The kinematics of the knee and ankle were calculated using Euler decomposition\(^7\) with rotation first about the flexion/extension axis, then the long axis of the bone, and finally the adduction/abduction axis. Total 3D translation
of the tibia was measured from the most distal, medial and anterior position of the tibia relative to the femur. The FHA of the rotation of the tibia was calculated for each time step and the average FHA position for stance phase was calculated to give an estimate of the change in the motion. In addition, the pattern of the interception of the FHA with the mid-sagittal plane (z=0) over the entire stance phase was evaluated, to determine the path of the center of rotation over the entire motion.

**RESULTS:**

The knee flexion-extension (Fig. 1a) curve for the AFO was more similar in shape to normal than the KO curve. However both displayed large differences at specific points particularly at weight acceptance (10-30% stance) and push off (100% stance). Differences between conditions were more clearly seen in the internal/external rotation of the tibia (Fig. 1b) where the range of rotation during the AFO trials (mean 24±3° std. dev.) exceeded that of the normals (12±10°), while range of the KO trials (5±1°) was less than normal.

For all rotations, the FHA was oriented primarily in the medial/lateral direction (rotation in the sagittal plane). The main effect of bracing was to change the angle of the axis in the frontal plane, with minimal effect in the transverse plane. During AFO trials, the average FHA was rotated 35±25° in the frontal plane. This was larger than during the KO (18±14°) and normal (8±23°) conditions. For normal subjects, the intersection of the FHA with the mid-sagittal plane during stance remained anterior to the midline of the femoral coordinate system for the first two-thirds of stance. The FHA then moved posteriorly at terminal stance, with the intercept at take off posterior and superior to the origin of the femoral coordinate system. For the AFO and KO conditions, take off occurred much closer to the origin of the femoral coordinate system. The pattern of intersections for the AFO trials is more similar to those of normal stance than the KO trials. However, the shift of the FHA anteriorly during early stance in the AFO trials is quite different than that seen in the normal subjects. The KO demonstrated a more constrained motion with the FHA located around the femoral origin for the majority of stance.

**Figure 1:** Knee kinematics: Normal data indicated by the gray band, AFO data by the green band and FKO data by the purple band. Time is normalized to percent of stance, 0% touch down and 100% take off. The band of data represents the mean of all trials ± one standard deviation.
For knee translations, the curves for the affected subject in the braced conditions were similar to each other and different from the normals in anterior-posterior (AP) translation (Fig. 1d), although the normals were more variable. At the same time, the range of the AP translation in the KO was reduced considerably (11±3 mm compared to the AFO and the normal conditions (19±4 mm and 20±8 mm respectively). The magnitude and variability of axial knee translation (Fig. 1e) were reduced for both the KO (2±1 mm) and the AFO (4±2 mm) conditions relative to the normal condition (21±11 mm). For medial/lateral translations (Fig. 1f), the normals showed little deviation during stance, while with the AFO, the tibia moved laterally in initial and terminal stance, and during terminal stance only with the KO. The total 3D magnitude of tibial translation at the knee joint for the AFO trials (17±4 mm) was closer to the range seen in normal subjects (13±10 mm) but was reduced with the KO (7±3 mm). Neither of the AFO or KO total 3D magnitude of tibial translation curves were similar in shape to the normals.

For the GRF, the braced conditions differed from normals in the vertical and AP components. In these plots, loading decreased in the braced conditions, particularly during weight acceptance and push off (approx. 20% and 80% of gait cycle respectively). In addition, small differences in medial/lateral forces were seen at initial and terminal stance for the braced conditions.

DISCUSSION:
The kinematics of the AFO and KO conditions differed from the normal trials. When using the KO, the subject had an average FHA closer to normal values and the overall 3D-tibial translation was reduced compared to the AFO. However, the range of the 3D-tibial translations with the AFO was closer to the normals than with the KO. The subject preferred the AFO due to a perceived increased stability compared to the KO, but from the kinematics, no clear improvement of the AFO over the KO is seen. In fact, from the FHA results, the KO appears to be have a more stable axis of rotation than both the AFO and normal conditions.

Tibio-femoral braces are often used in the treatment of knee instability, but it is not intuitive that an ankle brace would be useful in treating the knee. It is interesting to note that the AFO does not appear to constrain the ankle of the subject more than using the KO. Perhaps when wearing the AFO to constrain the ankle the subject was able to use a neuromotor normal strategy that could focus on knee stability. A more constrained knee is one reason a KO is worn for an ACL-deficient knee, but this must be considered relative to the subjective comfort and support allowed by the flexibility at the knee when wearing an AFO. If the AFO reduced quadriceps activation during gait, the subjective preference for the AFO may be linked to changes in muscle activation timings and the decreased quadriceps and increased hamstring activation seen previously in ACL-deficient subjects. Similar changes in muscle activation during gait are described by Berchuck et al. as a “quadriceps avoidance” pattern proposed to reduce knee moments in the ACL-deficient knee. To better evaluate differences related to the AFO and the KO, muscle activation patterns and joint moments must be evaluated, in addition to kinematics. Further investigation with more subjects and a measure of muscle activation is required.

In conclusion, the hypothesized difference in kinematics of the subject walking with the KO and AFO were observed, but it was not clear that the kinematics in the AFO were closer to normal values. This is one of a few studies that measures the 3D rotations and translations at the knee in an KO, and the first to do so in a patient wearing an AFO. In addition, the FHA method is presented as another way to evaluate and compare kinematics between different bracing conditions, trials or subjects. This case study highlights the need for further study to determine which brace is best for ACL-deficient patients and outlines some techniques to allow this to happen.