Three-Dimensional *in vivo* Motion Analysis of Knee Joint Laxity under Torsional Loading

Andrea Hemmerich [1], Willem van der Merwe [2], Christopher L. Vaughan [1]
[1] Department of Human Biology
University of Cape Town
Cape Town, South Africa
[2] Sports Science Orthopaedics Clinic
Sports Science Institute of South Africa
Cape Town, South Africa
ahemmer@cormack.uct.ac.za

Abstract— Excessive knee joint laxity is often used as an indicator of joint disease or injury such as rupture of the anterior cruciate ligament (ACL). Although the ACL has been shown to contribute to rotational stability of the knee, clinical assessment devices are currently limited to anterior-posterior drawer measurements.

The objective of this work was to design a methodology whereby *in vivo* knee joint kinematics could be evaluated in three dimensions using magnetic resonance imaging (MRI) to eliminate soft tissue artefact while still maintaining a non-invasive procedure.

An MRI compatible device is securely fastened to the outer frame of the open MR magnet and has been designed to administer a torque in the transverse plane while the subject's knee is being imaged. The rotational torque is applied manually to the foot and shank; the magnitude is measured using a strain-gauge. Additional rotation in the coronal and sagittal planes (up to 15° of abduction-adduction and 80° of flexion) is also permitted. When the predetermined torque has been reached, the foot is clamped in place and the strain gauge is disconnected prior to imaging, preventing image distortion.

The 3D T1-weighted sequence of the commercial 0.2 Tesla MRI generates 50 contiguous slices of less than 2 mm thickness (256 × 256 matrix). This 3D image volume is acquired in less than 5 minutes.

Segment coordinate systems are defined for both the femur and tibia based on the 3D positions of easily identifiable landmarks on the MR images. 3D rotations and translations of the tibia with respect to the femur are calculated by comparing the transformation matrices before and after torque are applied.

This system used to measure *in vivo* non-weight-bearing knee joint kinematics is non-invasive, avoids soft tissue artefact during the 3D measurements, and permits soft tissue visualization beneficial in clinical assessment.

Keywords— knee joint laxity; magnetic resonance imaging (MRI); three-dimensional (3D) kinematics; measurement system; anterior cruciate ligament (ACL)

I. Introduction

Disease of the musculoskeletal system can often be diagnosed by measuring stability of the affected joint. Excessive laxity is typically an indicator of knee injury, such as rupture of the anterior cruciate ligament (ACL). Devices used to diagnose ACL injury are currently limited to measurement of anterior-posterior (AP) translation since this is the direction in which the most severe laxity is observed. Measurement of rotational laxity of the joint is therefore simply based on the subjective assessment of the clinician.

Significant rotational laxity was shown to follow ACL rupture in porcine knees [1]. It has also been shown, using cadaver knees, that both a more anatomical tunnel placement [2] as well as replacing both, instead of just a single bundle of the ACL [3], restores rotational stability. These studies were limited, however, in that they were unable to examine human *in vivo* results due to the invasiveness of the methodologies.

Several studies have been conducted using imaging techniques such as magnetic resonance imaging (MRI) or stress radiography in order to examine the underlying bones *in vivo*. Lerat *et al.* [4] and Logan *et al.* [5] observed rotational instability of the knee during the Lachman test, which administers an anterior-posterior stress. Further studies have examined joint laxity under valgus stress [6]; however, little data is available on knee joint laxity under torsional loading.

In order to determine the function of the ACL in the transverse plane, a reliable method of measuring the kinematics is required. The objective of this work was to design a methodology whereby *in vivo* knee joint kinematics could be evaluated in three dimensions using MRI to view the underlying bone and soft tissues and to eliminate soft tissue artefact. This non-invasive procedure, using a specially designed knee torque device, is intended to compare rotational knee laxity following ACL injury and surgical reconstruction to normal knee kinematics.

II. Design of 3D Knee Loading Device

A. Device Kinematics

The open-MRI compatible knee loading device was designed to apply stresses that simulate those applied by a clinician during injury assessment. Accordingly, rotations in all three planes (sagittal, coronal, and transverse) are permitted. The knee angle is determined using the device by positioning the coupled foot and shank relative to the thigh which remains fixed. The foot is rigidly connected to the shank by way of a plastic air cast. Up to 15° of varus-valgus movement, in addition to 80° of flexion can be accommodated; internal and external rotation is limited only by interference between the MRI patient table and foot, depending on the position of the subject.

The subject may lie supine or prone when the knee is loaded from full extension to 10° of flexion. For greater flexion angles, the subject is required to lie on the same side as the knee being imaged. Fig. 1 shows the knee loading device mounted to the MRI patient table for imaging of the right knee. The slide rails permitting flexion-extension and abduction-adduction, as well as the disc for internal-external rotation are shown. The air cast is connected to the rotation base via extension channels that permit foot positioning toward the knee coil for shorter subjects as shown in Fig. 2. The knee loading device is rotated about the patient table to permit imaging of the contralateral knee.

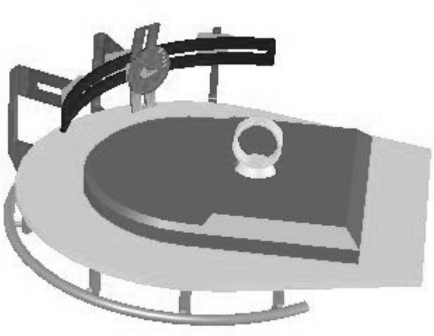

Figure 1 The 3D MRI-compatible knee loading device mounted to the patient table. The slide rails and rotation disc permit knee rotation in the three planes.

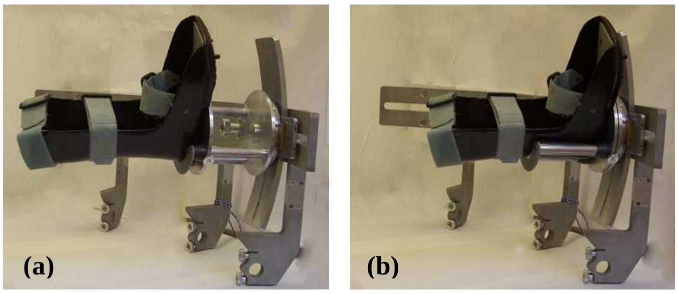

Figure 2 Extension channels allow the foot plate to be positioned closer to the MRI knee coil. The air cast is shown in (a) the fully extended position and (b) the fully contracted position.

This work was financially supported in part by the Natural Sciences and Engineering Research Council of Canada (NSERC).

Stresses are applied to the distal end of the foot and shank which are rigidly connected in the air cast. The applied torque is therefore transmitted to the knee joint via the tibia.

1) Magnitude of Applied Stresses

In order to compare results between individuals and subject groups, a set torque is applied to each knee being examined. However, in order to account for the subject's weight which regularly affects the loads experienced by the knee, the applied torque is normalized to body mass.

The magnitude of the applied torque is measured using a strain gauge mounted onto a load cell. Fig. 3 shows an exploded view of the main components of the torque mechanism. An internal torque (illustrated by the curved arrow) is manually applied to the torque disc, which transmits a linear force (illustrated by the vertical arrow) to one end of the load cell. The load cell is rigidly fixed to the rotation base and, in turn, the foot which is resisting the applied torque. The resulting strain deformation in the load cell is measured by the strain gauge with data collected and displayed on the user interface using LabVIEW™ (National Instruments) software. External torque is measured by a second strain gauge mounted onto the opposite side of the load cell. Calibration of each side of the load cell was achieved by hanging weights off the end of a lever arm of known length extending from the torque disc while the rotation base was rigidly fixed in place.

By manually applying the torque in the specified direction using the 3D knee loading device, the investigator is able to determine an appropriate rate of torque application to ensure comfort of the subject. Once the correct torque has been reached the foot, connected to the device, is clamped in place to maintain the applied load. The strain-gauge is then disconnected to prevent image distortion during MRI scanning.

III. Magnetic Resonance Image Acquisition

MR images were acquired using the 0.2 Tesla dedicated open-MR system (E-Scan XQ, Esaote, Italy). A 3D T1-weighted sequence with a 256 × 256 matrix was used to generate approximately 50 contiguous slices of less than 2 mm thickness for a 10 cm field-of-view. This 3D image volume is acquired in less than 5 minutes.

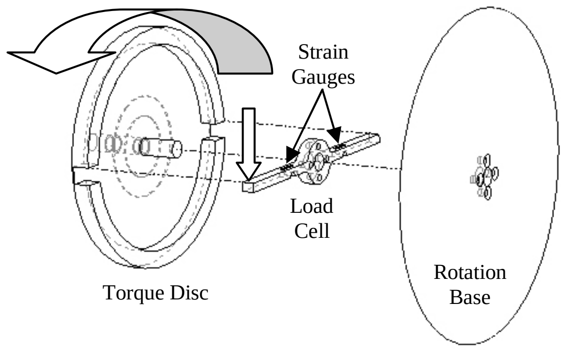

Figure 3 Exploded view of the components of the torque mechanism.

IV. 3D Kinematic Description of Knee Motion

A. Segment Coordinate Systems

By identifying several bony landmarks on the MR images of the femur and tibia, a local coordinate system (LCS) for each bone was defined (see Fig. 4). Once the landmark has been identified in all three planes, its 3D position data is exported into MATLAB® to calculate the LCS.

The y-axis of the left femur extends from the medial to the lateral femoral epicondyle with the origin at its midpoint. The femoral x-axis in the posterior-anterior direction is perpendicular to the y-axis and a temporary z-axis. This temporary z-axis is normal to the plane defined by the most anterior and posterior points of the medial femoral condyle and the most posterior point on the lateral condyle. The final femoral x-axis is the cross-product of the x- and y-axes.

The origin of the left tibial coordinate system is approximately in the middle of the medial plateau, since the axis of rotation extends through this position for the flexion range of 10º - 80º [7]. This location was defined as 0.2 times the medial to lateral condyle distance on the y-axis. The tibial y-axis extends from the most prominent points on the medial to lateral tibial condyles. A temporary z-axis used to calculate the x-axis is defined as normal to the plane consisting of the most medial and lateral points on the medial and lateral condyles, respectively, as well as the most anterior point on the medial condyle. The x-axis of the tibia is the cross-product of the y- and temporary z-axes and the final tibial z-axis is the cross-product of the x- and y-axes.

The transformation matrix relating the 3D position and orientation of the tibial coordinate system to the femoral coordinate system was calculated before and after loading. The transformation matrix relating the 'before' and 'after' positions describes the six degree-of-freedom motion that has occurred due to loading.

B. Application to the Joint Coordinate System

Rotation and translation measurements follow the convention developed by Grood and Suntay [8]. The flexion-extension axis is defined as the x-axis of the femoral coordinate system, the internal-external rotation axis is defined as the tibial z-axis, and abduction-adduction occurs about the floating axis which is perpendicular to the preceding two axes.

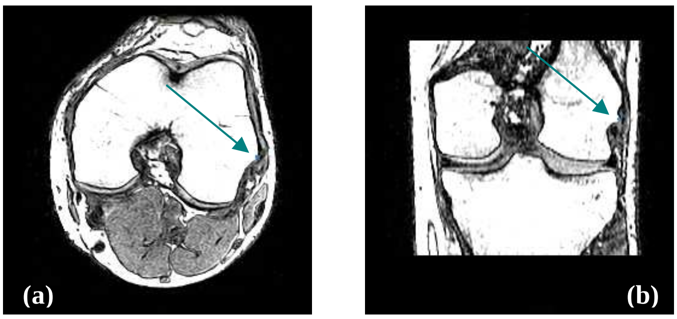

Figure 4 Three-dimensional transverse (a) and coronal (b) plane MR images showing the segmented medial femoral epicondyle.

V. Validation

The device was tested to ensure that it would not cause image artefact during MR scanning. The accuracy to which relevant landmarks can be identified on the MR images at various flexion and rotation angles is within 2 mm. A mechanical knee model was used to determine the accuracy to which the device can position the knee joint by measuring rotation in the two primary planes of motion independently of the loading device. The tibial component of the knee model was rigidly connected to the foot plate of the device and positioned randomly at various rotation angles; its position relative to the femoral component of the knee model was ascertained from the position of the kinematic components of the device. Preliminary results have shown an accuracy of device rotation of better than 2° in the flexion-extension plane and 1° in the internal-external rotation plane.

Further validation is currently being undertaken to determine the accuracy and repeatability to which knee translation and rotation can be measured from the MR images. Positions of the tibial and femoral components of the knee model are measured from the device and compared to those measured from MRI scans using contrast medium for landmark identification. A further study to ensure that results from a single subject are repeatable for both left and right knees at various flexion angles will be conducted. Device validation, as well as pilot test data collection and processing will be completed by June, 2006.

VI. Limitations of Design

There are limitations to the proposed system that may influence the accuracy of the results obtained from its measurements. Since the device will be used without administering any anaesthetic to the subject, the applied stress may cause muscle contraction, thereby reducing the measured joint laxity. By making sure the subject is positioned comfortably and by reducing the rate of torque application as necessary, the investigator will be able to ensure that the muscles of the knee are as relaxed as possible.

The accuracy with which the bony landmarks can be identified on the MRI in 3D space depends on the slice thickness and resolution of the image; however, if resolution is increased, scanning time also increases. Current scan sequences produce results whose accuracy is comparable to that of the landmark identification process.

Landmark identification on the MR images is currently performed manually, since existing automated segmentation techniques are subject to error. Although a laborious process, manually marking the bony landmarks on the MR images requires less than 30 minutes per subject, since complete segmentation of all the tissues is not required.

VII. Conclusion

An innovative methodology used to measure 3D laxity of the knee joint under torsional loading has been demonstrated. The system consists of a device designed to apply a predetermined torque to the knee joint and a method of measuring the resulting knee kinematics by means of medical image analysis. The advantages of this technique include the use of MRI which is non-invasive and permits accurate measurement of the underlying bone, thereby avoiding skin motion artefact. Furthermore, MRI allows the visualization of soft tissues around the joint; injury to these tissues may best be seen under loading.

ACKNOWLEDGMENTS

The authors would like to acknowledge Sarah Dawson, Charles Harris, Iekraam Fakier, Melissa Dagnin, and Merle Neethling du Toit for their work on the device design and data collection.

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