Introduction
Surgical intervention of the knee joint routinely endeavors to recreate a physiologically normal joint loading environment. The in vivo loading conditions are known to be dependent on the interplay of muscle activity and the kinetics of the knee joint [2]. Open MRI based procedures allow assessment of knee kinematics non-invasively and with a high degree of precision [2]. A number of computational models have been proposed to describe knee kinematics [3, 4], but have been limited to 2D analysis [3], describe passive motion only [5], or lack validation against in vivo data from a larger population [4]. Whilst the four bar linkage mechanism has been proposed by several authors as a model to describe the knee kinematics [3, 6], this model has also been criticized as oversimplifying the complex three-dimensional in vivo kinematics [7]. The goal of the current study was to determine whether a new, three-dimensional model of the knee kinematics that modifies the four bar linkage could predict the kinematics of the femoro-tibial joints and the loading of the ligaments in the knee during unloaded and loaded knee flexion.

Materials and Methods
The knee joints (no history of pain or injury) of 12 healthy volunteers were examined. Kinematic analysis was performed in an open MR system (0.2T: Magnetom-Open, Siemens, Erlangen, Germany) using a T1-weighted 3D gradient echo sequence (TR 16.1ms, TE 7.0ms, FA 30°). Sagittal plane images (slice thickness: 1.875mm, in plane resolution: 0.86mm) were taken with an acquisition time of 4min 26s. The volunteers were placed on their side and the knee flexion angle (0°, 30°, 90°) was controlled by a special positioning device (Siemens, Erlangen, Germany). Scans were taken with the knee either unloaded or with the application of a weight of 3kg to the lower third of the shank to produce a torque of 10Nm. This torque was initiated in a flexion direction which led to activation of the extensors. Isometric muscle activity over the entire acquisition period was verified by surface electromyography. All parts of the study were approved by the local ethics committee, and written consent was obtained from all participants prior to MR imaging. After image acquisition, semi-automated segmentation of the femur, tibia, ACL and PCL was performed based on a grey-value oriented region-growing algorithm. Surface models of all structures were then created using a marching cubes algorithm.

The kinematic model deviated from previous four bar linkage models in that all 3D ligament attachment points and each patient’s individual internal-external (I-E) rotation served as an input to the model. In addition, the lengths of the ligaments were not constrained to be isometric but were allowed to vary within a range of min and max strain [8]. The model then set out to determine a set of parameters (attachment positions and ligament strain) that minimized the deviation between the pose of the femur predicted by the kinematic model and the one determined directly from the in vivo MR data. After registering the tibial surfaces at 30° and 90° to the 0° flexion datum, used as a reference for all bone and ligament surfaces, bone coordinate systems were defined to derive the in vivo kinematics. In vivo I-E rotation was calculated with reference to a trans-epicondylar axis [2]. The bony attachment areas of the ACL and PCL were fitted by ellipses onto which grids were imposed, defining a total of 59 points for each attachment site. The kinematics of the joint, and the resulting bone poses were then predicted under each possible four bar linkage, constructed from combinations of attachment points. From all possible solutions, the combination of points that minimized the sum of the distances between three reference points on the femur for the pose predicted by the kinematic model and the in vivo measured pose was selected. A similar solution was sought for the case when no I-E rotation was included. The finally achieved match between the predicted and the actual in vivo pose of the femur was established by computing the Hausdorff distances between the 0° datum and the surfaces reconstructed directly from the 30° and 90° MR data. Ligament strains throughout the full flexion cycle were computed as the distance between the femoral and tibial attachments divided by the length of the 0° flexion, based on the prediction of the kinematic models, and compared against the in vivo data at 30° and 90° flexion.

Results
The deviation between the modeled and the in vivo pose of the femur clearly decreased when I-E rotation was included. The average distance between the reference points at 90° was reduced from 12.5±6.3mm to 6.6±3.4mm for the unloaded and from an average of 11.3±5.6mm to 7.8±3.6mm for the loaded knees. Whilst the reference points were chosen at locations where the largest errors were expected, the Hausdorff distances at 90°, which averaged 1.1±0.6mm and 1.4±0.7mm for the unloaded and the loaded knees respectively, indicated that the overall match between predicted and in vivo pose of the femur was good when the I-E rotation was included. Furthermore, when I-E rotation was included, the largest difference between the ligament length according to the model and the in vivo data was only 3.1% for the ACL and 0.9% for the PCL. Whilst the kinematic model predicted the ACL to become less tight, PCL strains were increasing with flexion (Figure 1). The kinematic model predicted isometric behavior of the AM bundle of the ACL for flexion angles below 30° when unloaded, whilst for the loaded knee this behavior was found throughout flexion. Shortening of the PL bundle of the ACL was more pronounced under loaded than unloaded conditions. The largest strains in the PCL were observed in the loaded AM bundle (126%).

Discussion
This study has validated a new 3D model of human knee kinematics against in vivo data for passive motion, but more importantly, for physiologic neuromuscular activation patterns in 12 healthy subjects. By extending the four bar linkage model, i.e. by allowing for non-isometric fiber lengths of the cruciate ligaments and by including I-E rotation, it was possible to predict the kinematics, as well as the ligaments strains, throughout a full cycle of knee flexion. Not only did the kinematics and the loading agree well with the in vivo data from the same patients, but the loading of the different bundles of ligaments was also well in agreement with the findings of recent studies [9]. It seems likely that by integrating such patient specific models of the knee kinematics with musculoskeletal analyses of the lower extremities, a better understanding of the load distribution within the structures of the knee can be obtained. Future research should focus on the specific conditions in patients with e.g. focal osteoarthritis at the knee, and may provide useful information to guide the development of therapies.

References

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