Chapter 6

General discussion
Providing a patient-specific 3D knee finite element (FE) model is one of the proposed solutions to optimize the ACL reconstruction result. An ACL reconstruction is the common solution to treat an ACL injury and to obtain a well-functioning knee (Beynnon et al., 2002). By using FE modeling, we can simulate and subsequently analyze the effects of important parameters that are influencing the surgical outcome and which are difficult to assess in experiments (Peña et al., 2006) or during surgery.

Before performing these biomechanical computer simulations, the FE models need to be validated. The reliability of the knee model is strongly depended on an appropriate geometrical reconstruction, realistic mechanical behavior of the surrounding soft tissues (Peña et al., 2006) and adequate boundary conditions of the model (i.e. kinematics and muscle forces) (Besier et al., 2005). The gold standard used to validate FE models may be an experimental (cadaveric) assessment. In that case, the quality of the FE model simulation result becomes directly dependent on the quality of the experimental data which is obtained from the experiment (Beillas et al., 2004).

An essential step to reconstruct a cadaver or patient-specific FE knee model is to develop a 3D knee joint geometrical model. Some factors should be considered during the model development i.e. the time, accuracy and costs and these factors are not independent from each other. The development time includes the scanning time of the knee to have a better digital image quality. Subsequently, the segmentation procedure to reconstruct the 3D geometrical model from the image is usually labor (and time) intensive. In addition, the accuracy of the model is associated with the accuracy of both the geometry and material properties of the knee bones and soft tissue structures. The costs include the costs for the scanning, experimental assessments and model development.

For example, the obtained accuracy of the geometry will be influenced by the scanning sequence. By using other (more time consuming) scanning sequences it is possible to increase the image contrast for each knee structure. Furthermore, different image sequences can be combined (fused) to improve the noise-to-signal ratio of different types of structures (e.g. for the bone a different optimal sequence may be used as compared to the meniscus tissue). A better image quality will increase the visibility and shape (resolution) of the knee structure. Hence, improved image quality will lead to a better and faster segmentation of the knee tissues.

Based on the application of our study, we still used a CT and an MRI scanner with a standard clinical scanning sequence (Chapter 2) to scan the full knee bones structures (femur, tibia, fibula and patella) and the short knee bones (around 20 cm upper and below knee joint cavity) and soft tissues. The scanning time of the CT and MRI was also still reasonable: 6.5 minutes (2200 slices) and 12 minutes (230 slices) on average for the CT and MRI scans, respectively. The scanning yielded the images with an adequate contrast and provided enough information of the bone structures to be segmented. In this thesis we show that we have been successful to develop the knee geometrical model which consists of bone and soft tissues structure. We were also successful to
develop the bone geometrical models with a good quality and the same knee bones volume models from two different images sources i.e. MRI and CT.

By using thin slices of the MRI and CT images, we found that we can develop a smoother 3D knee geometrical model. However, we found that the process of segmentation was more time consuming. We required more time to segment, especially, the bones structures from the CT scan images, due to the high number of image slices which had to be manually segmented. We also scanned with two different scanning modalities (CT and MRI) which increased the required time as well, but allowed for the creation of a knee model which contained the soft tissue structures (MRI-based) and the density distributions within the bones (CT-based).

By using a clinical scanning sequence, we had great difficulties to recognize the soft tissue structure in detail (e.g. for the ligaments: shape and attachment sites) which greatly hampered the segmentation process. This problem will have a significant impact on the segmentation time and obtained quality of the segmented tissues. Hence, it is critical to improve the quality of the MRI scans regarding of the soft tissue images if we were to generate truly patient specific knee models.

Because we found that we could obtain bone geometry from the MRI scans in an accurate manner, we propose to employ only MRI images as a digital image source to develop the knee model. An additional advantage is that the MRI scan is safer for the patient as compared to the CT scan from a radiation exposure point of view. The MRI image quality can probably be further improved by scanning the knee using a circular polarized extremity coil in supine and extension position (leg externally rotated 10° to 15°). Another scanning MRI sequence i.e. T1- weighted spin-echo sequence (TR 300 ms, TE 30 ms, 5 slices, slice thickness 3.0 mm, gap size 0.5 mm, matrix 256 x 360) can be used to scan the knee in order to improve upon the image quality of the knee soft tissue anatomy, especially ACL, PCL and intercondylar roof. This MRI scan protocol has been successfully applied by Staebul and colleagues (Staeubli et al., 1999) to assess the anatomy of the intact ACL, PCL and femoral intercondylar notch on cryosections of a knee cadaver specimen in the coronal oblique plane oriented parallel to the intercondylar roof.

Another requirement to improve upon the segmentations of knee anatomical structures is a higher level of knowledge to recognize the tissues on the MRI images. Obviously training with an experienced radiologist would be optimal but it is important to realize that the researcher has to train him/herself in segmenting these complex structures. It would therefore be recommendable to perform training sessions with an experienced radiologist on images of cadaver knees which are also anatomically dissected to enhance the understanding of the complexity of the knee structures around the knee joint.

One of our studies evaluated the intra and inter variability of the identification of knee ligament attachment sites in MRI scans (Chapter 3). Accurate identification of these sites are very crucial in order to predict the mechanical behavior of the particular knee. We involved five observers and compared the results with experimental measurements.
Previous studies indicated substantial implications when malpositioning the attachment sites of the ACL (Bedi et al., 2010; Howell, 1998; Zavras et al., 2005), PCL (Galloway et al., 1996; Petersen et al., 1996; Oakes et al., 2003; Shearn et al., 2004; Mannor et al., 2000; Burns et al., 1995), MCL (medial collateral ligament) (Bartel et al., 1977) and LCL (lateral collateral ligament) (Meister et al., 2000). We propose to perform more comparative studies using more (cadaver knees) to train researchers and computer algorithms to identify true attachment sites. This information can be combined with the improved semi- or full automated segmentation procedures. If the modeling tools are to be applied in a clinical setting these types of (semi)automated segmentation methods become essential.

To provide more insights in the ligament fiber orientation in combination with the identification of the insertion sites, other MRI-based techniques such as a diffusion tensor imaging can be used, which possibly could be used to identify the fibre directions of the ligaments (Froeling et al., 2012). Furthermore, the attachment sites could also be found in a more robust manner if multiple MRI scans are made in different joint (flexion) angles. By extrapolating the direction of the ligament from different angles, the attachment sites positions could be identified in a more unique manner thereby reducing the intra and inter variability. Obviously, this type of scanning will increase time and costs and it remains to be proven that this method indeed improves accuracy of the methods. The live wire method is one example of semi-automatic segmentation method which introduced by Barrett and Mortensen (1997) and implemented by Zheng et al. (Zheng et al., 2011) to segment lumbar vertebrae. This method seems also suitable to segment the soft tissue structures from the MRI images.

Another approach for semi-automatic segmentation is the integrated mesh-morphing approach which can be used to segment bone cartilage (femoral, tibial and patellar) and patella bone and has been published by Baldwin et al. (Baldwin et al., 2010). Their study proved that their segmentation could create hex-meshes of knee soft tissue structures and bone structure from MRI images with an average geometric difference of 0.54 mm (root mean square) and a peak difference of contact pressure of less than 5.3% (quasi-static analyses over a range of flexion angles) between semi-automated and a manually developed model.

Another semi-automatic segmentation has been implemented by Swanson et al. (Swanson et al., 2010) to segment meniscus tissue. They segmented the meniscus by first seeding the meniscus area manually and followed to calculate thresholding level through a Gaussian fit model. Their segmentation process was ended by conditional dilation and post-processing under anatomical, intensity and range constrains. Their study showed that the segmentation produced accurate and consistent segmentations of the meniscus.

In order to reduce the time of model development, after extracting many meshes of the knee 3D geometrical structures, a database of aligned surface knee model structures can be built. According to a study published by Schuman et al., the database development is the first step to use a statistical shape model (SSM) (Schumann et
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al., 2010). By using the SSM method, we can possibly develop the subject-specific 3D surface models from some or limited 2D images such as X-ray or fluoroscopy. By extracting patient-specific information from 2D images, the morphology of the knee in the database can be statically analyzed and used to compute a new shape and create a new 3-D model. The variability in the database is computed using principal component analysis (PCA).

The 2D/3D reconstruction method has been successfully used to reconstruct 3D geometries from 2D images such as a proximal femur (Zheng and Schumann, 2009; Schumann et al., 2010; Boussaid et al., 2011), bone cartilages (femoral, tibial and patellar) and patella bone (Baldwin et al., 2010), pelvis (Zheng, 2010); hip joint (Schumann et al., 2013) and lumbar vertebrae (Zheng et al., 2011). We propose to use this method for future application to develop bones and soft tissue structures of the knee. The method could also be used in a semi-3D manner in which some slices of the MRI images are required to fit the statistical shape model. Hence, statistical shape models can be applied using either 2-D or sparse 3-D imaging information. An obvious disadvantage of this method is that the modes of variability of the structures need to be incorporated in the training data set. Hence, severe pathological (shape) cases will not be captured by this type of segmentation unless they are included in the training set.

The consideration to include the posterior knee capsule (Chapter 4) to the knee joint model is the next things to improve in order to have a model closely mimicking the native knee. We therefore considered the role of the posterior knee capsule, which influences the biomechanics of the knee joint especially at full knee extension (Moeizadeh and Engin, 1983). The calculations in our study estimated that the posterior knee capsule produced a force about more than half a body weight during 10 degree of hyperextension of the knee. Hence, we have shown that the posterior knee capsule may considerably affect knee biomechanics. The absence of this structure in the biomechanical models will not only create lower contact forces at full extension, but also triggers errors in other structures which (partly) take over the function of the posterior capsule at full flexion. How large these errors are remains to be seen and biomechanical models (in combination with experimental measurements) can be used to provide more insights in this.

Simulating the knee capsule is highly challenging. The anterior and side parts of the capsule are very narrow, whereas the posterior part is wider (which is the reason that we selected the posterior part of the capsule as we could obtain test specimens). In the anterior part, the knee joint capsule attaches to the quadriceps and patellar tendons (Ralphs and Benjamin, 1994); on the medial and lateral sides, the capsule is blended with the ligament and the meniscus. The knee capsule is a highly complex structure, with non-homogeneous material constituents and a varying fiber orientation and distribution. Therefore, further mechanical tests are required to obtain the mechanical properties.

To understand the complex behavior of the knee and its soft tissue better one could also opt to use (dynamic) MRI imaging. More detailed measurements relative to
knee joint kinematics such as while performing a daily activity can be further performed using the dynamic MRI. For instance, dynamic MRI is feasible to examine abnormal knee joint mechanics due to an MCL injury (Studler et al., 2011) and to evaluate in-vivo load-dependent variations in tibio femoral and patella femoral kinematic (Westphal et al., 2013). Using dynamic MRI a knee could be loaded (e.g. in varus or valgus) and the strains could be assessed from the images. These experimental conditions could be simulated with the FE models so as to tune the mechanical properties of the soft tissues in the model. These experimental-simulation combinations could also be used to define the slack lengths of the various ligaments as we know that this parameter is considerable affecting the biomechanical behavior of the knee.

By using a validated knee joint model, the surgeon can use it to simulate the important parameters for achieving the most optimal ACL reconstruction. The simulation possibly gives a description of the knee kinematics after surgery and the force distribution during the physical activities. Furthermore, it could assist in determining the optimal rehabilitation program for patient after surgery. We believe that with the current ACL reconstruction techniques, the cartilage will have a different stress pattern than physiologically and the challenge of the reconstructive surgery is to generate kinematics, strain and stress patterns which are as physiological as possible. With a patient specific imaging-modeling approach as described in this thesis that goal becomes more feasible, although there is a long road ahead to apply these techniques in the clinic.

In summary, by performing the studies as described in this thesis, we have made a step forwards to develop tools to simulate ACL injured patients in a patient-specific manner. We have added knowledge about the posterior knee capsule which is important at full extension and assessed ACL behavior in-situ. It is clear that there is still a long way to go in terms of imaging, segmentation and determination of mechanical properties, before actual patients can be simulated in a fast and accurate manner.

References


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