Towards a subject-specific knee model to optimize ACL reconstruction
Rachmat, Hendi

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Chapter 2

Development of a patient-specific 3D knee joint geometrical model to optimize anterior cruciate ligament (ACL) reconstruction

H.H. Rachmat¹³⁶, D. Janssen³, R.L. Diercks², G.J. Verkerke⁴⁵, N. Verdonschot³⁴

¹ Department of Biomedical Engineering, University of Groningen, University Medical Center Groningen, A. Deusinglaan 1, 9713 AV Groningen, The Netherlands
² Department of Orthopaedics, University of Groningen, University Medical Center Groningen, The Netherlands
³ Orthopaedic Research Laboratory, Radboud University Nijmegen Medical Centre, P.O. Box 9101, 6500 HB Nijmegen, The Netherlands
⁴ Department of Biomechanical Engineering, University of Twente, Enschede, The Netherlands
⁵ Department of Rehabilitation Medicine, University of Groningen, University Medical Center Groningen, The Netherlands
⁶ Department of Electrical Engineering, Institut Teknologi Nasional (ITENAS) Bandung, West Java, Indonesia.
Abstract

Three dimensional subject-specific finite element (FE) models of the knee joint can be developed to optimize the outcome of ACL reconstructions. By using FE modelling techniques, parameters such as the optimal location and length of the new graft can be studied before surgery in order to achieve a successful patient-specific ACL-reconstruction. One important aspect in this procedure is to have an appropriate 3D knee model. The goal of the current study was to develop 3D knee models from CT and MRI scan images. By using reconstructed models, we compared the CT-based and MRI-based 3D knee bone models to evaluate the quality of the geometrical representation. Furthermore, we evaluated the volumes of bony and soft-tissue segments to assess whether knee models could be generated by rather simple volumetric scaling methods.

By using five right knees from five cadavers, we successfully reconstructed the bony segments based on the CT images. We furthermore reconstructed the bony segments and knee ligaments based on the MRI data. We succeeded to register CT based knee bone models to MRI based knee models using the bone structures (maximum error of 1.02 mm for long bones and 2.05 mm for patella bone) for developing a combined knee joint model. We found that the volume of the MRI bone structures was comparable to the CT bone structures with a maximum volume difference of 6.64% for long bones. We could not find a good scalable (volume) factor for all structures amongst the different cadaver knee models, particularly for the soft tissue geometries.

These results indicate that 3D FE model of the knee for ACL reconstruction simulation can be developed by only using MRI scan images. However, further improvements are required in order to speed up the reconstruction process, due to the uniqueness of the knee structures from one patient to the other.

Key words:

Anterior cruciate ligament, geometrical model, knee joint, patient-specific.
1. Introduction

Of all knee ligaments, the anterior cruciate ligament (ACL) is the most frequently disrupted (Peña, et al., 2006) as a result of extreme pivoting and other abrupt changes in direction during sports activities. An untreated ACL-deficiency can provoke cartilage degeneration, leading to osteoarthritis and finally a total knee arthroplasty (Gao and Zheng, 2009; Andriacchi et al., 2006; Noyes et al., 1983; Nebelung and Wuschech, 2005). ACL-reconstruction is a common procedure to treat the disability or chronic instability of knees due to ACL injuries (Beynnon et al., 2002). It strives to return the patient to a previous level of functioning while preventing later degeneration of the knee (Cesar et al., 2008; Gao and Zheng, 2009). However, in several cases the reconstruction is not optimal, leading to cartilage damage. Our hypothesis is that this is due to the fact that the reconstruction is performed for the average patient, whereas the average patient does not exist and the biomechanical condition of a knee will be highly sensitive to even only small changes of a reconstructed ACL.

To solve this problem, we aim to develop a three-dimensional (3D) patient-specific finite element (FE) model of the knee joint. FE models have proven to be able to provide deep insights into the influence of mechanical properties of biological tissues on the performance of structures (Peña, et al., 2006). Through this model, it becomes possible to explore the important factors affecting successful ACL reconstructions, especially for determining the optimal location of the graft on femur and tibia, as well as the optimal length of the graft.

It is important to note that the reliability of these models strongly depends on an appropriate geometrical reconstruction and on accurate mathematical descriptions of the behavior of the biological tissues, such as ligaments, tendons and menisci and their interactions with the surrounding environment. An appropriately developed FE model is a powerful tool to predict the effects of parametric variations, and to provide information that is otherwise difficult to obtain from experiments (Peña, et al., 2006). To perform some passive knee movements using an FE subject-specific knee joint simulation, we need a 3D knee geometrical model of an intact knee including bones and soft tissues and material properties for each knee element.

These models are usually based on a 3D reconstruction of the knee joint from imaging data such as magnetic resonance imaging (MRI) and computerized tomography (CT). Some previous studies have used a set of CT scan images and MRI scan images to develop a geometrical model of the knee soft tissue elements and knee bones, respectively (Donahue et al., 2003, Peña et al., 2006, Suggs et al., 2003, Li et al., 1999, Potočnik et al., 2008). Some groups also reconstructed the geometrical model of the knee bones from the MRI scan images (Suggs et al., 2003, Li et al, 1999, Rathnayaka et al., 2012). MRI is safer for knee patients, because of the absence of ionizing radiation which is present with CT. However, CT is currently the gold standard for scanning of bones to develop 3D models with a high geometric accuracy (Rathnayaka et al., 2012). It furthermore has a shorter time of scanning, and it offers the ability to determine the bone density, which can be used to assign bone material properties to the FE model.
Therefore, in this study, we developed a subject-specific 3D knee model from a cadaveric specimen using CT images to develop 3D geometrical bone models and MRI images to develop 3D bones and soft tissue models. Besides to develop the geometrical knee model, the other goal of this study was to compare the reconstruction process of a subject specific 3D knee bone model between CT scan images and MRI scan images as well as to evaluate the scalable factor amongst the similar knee bones and soft tissue volumes from different cadaver knees based on MRI scan images.

2. Materials and methods

To develop the 3D geometrical model, this study utilized five right knees from five cadavers, which had an average age of 76 years (3 males, 2 females; range: 66-86 years).

2.1. MRI and CT scanning protocol

The legs were scanned using a Siemens 3 Tesla MRI scanner (Siemens, Trim Trio, Erlangen, Germany). We used an extremity knee coil to improve the contrast of all soft tissue elements in the digital images. The cadaver knee was scanned in the sagittal plane (Figure 1 (left)) resulting in 240 MRI image slices. The images were acquired using a clinical sequence according to the following parameters: spin echo (SE) scanning sequence with repetition time (TR) of 1200 ms, echo time (TE) of 29 ms, field of view (FoV) of 224 mm x 224 mm, matrix size of 448 x 448, pixel size of 0.5 mm x 0.5 mm and slice thickness of 0.5 mm.

Subsequently, the legs were scanned in the transversal plane (Figure 1 (right)) using a Siemens CT scanner (Siemens, Sensation 16, Erlangen, Germany) with scanning parameters as follows: 120 kVP, 330 mAs, matrix size of 512x512 and pixel size of 0.488 mm x 0.488 mm. The CT scans were made from the femoral head to the distal part of tibia. Due to an intermediate update of the software on the CT scan machine in the Radiological Department, we minimized the noise on the images by adjusting a slice thickness of the images scanning. The slice thickness values were 0.75 mm for one cadaver, 0.6 mm for two cadavers and 1 mm for two cadavers. The mean total slices of the CT scan images were 2243 slices (2243 ± 565 slices, range: 1858 – 3227 slices).

Figure 1. Sagittal view of an MRI scan of the knee joint in full extension (left) and a CT scan showing the femur and patella in the transversal plane (right).
The FoV of the images was not also the same, because of the different size of the legs. The mean FoV was 32.9 cm (range: 23.5 – 50 cm). During CT scanning, a calibration phantom was placed under the leg in order to enable future calculation of the bone mineral density (not used in this study).

All MRI and CT images were using in house developed software (DCM toolkit (dcmtk) software version 11.2, ORL St Radboud, Nijmegen, The Netherlands). The DICOM files were then imported in Mimics® 14.0 software (Materialise, Leuven, Belgium).

2.2. Reconstruction of MRI-based and CT-based knee models

The reconstruction procedure of the 3D knee models included the image segmentation and the 3D surface mesh model construction. Development of 3D knee surface models which consisting of four bone elements and soft tissues elements were performed using Mimics software. CT scans of one cadaver knee was used to reconstruct bone elements and MRI scans of five cadaver knees were used to reconstruct bone and soft tissue elements.

MRI-based knee models

After importing the images, the program created the MRI project file and presented the images on three different planes as shown in Figure 2.

![Figure 2. The three different planes of the MRI images as displayed in the Mimics software.](image)

Each part of the knee including bones and soft tissues was identified visually. We created two different main masks to define the bone and the soft tissue without using a thresholding method. The method was not working optimally for all MRI scan images. This was due to a different MRI Hounsfield unit (HU) scale of each cadaver.
Chapter 2

The next process was to separate each element from the bone mask as well as the soft tissue mask. The region growing tool was used to separate the thirteen knee elements by creating a separate mask for each element. We generated four different bone masks for tibia, femur, patella and fibula and nine different soft tissue masks including anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), lateral collateral ligament (LCL), medial collateral ligament (MCL), tibial articular cartilage, femoral articular cartilage, patellar articular cartilage, lateral meniscus and medial meniscus. Figure 3 shows the different masks for both bone elements and soft tissue elements of the knee model in the sagittal view. In each different region, we improved the quality of the masks by manual editing to add or to remove some pixels as well as to refine the edges for each slice to get the smoothest the 3D knee model.

![Figure 3. Example of the different mask regions in the knee model. In the left image, the femur (green), patella (pink), tibia (blue) and fibula (yellow) have been segmented, while in the right image the patellar (magenta) and femoral (green) cartilage, and the anterior (purple) and posterior (pink) cruciate ligaments have been segmented.](image)

Based on the best generated region masks, we stored each mask to create the 3D mesh surface model of the bone elements and soft tissue elements with a default medium quality setting. The mesh surface models were smoothed using the following parameters: smooth factor of 1 (range smooth factor from 0 until 1), 50 iterations, using shrinkage compensation. We performed the same procedures to reconstruct 3D models based on the MRI images. Figure 4 shows the 3D geometrical models from five different cadaver knees.

All elements were then exported as STL files for further processing. To evaluate the scalable parameter among the similar knee bones and soft tissues from different knees based on MRI scan images, the volume values were collected. The volume was calculated automatically by the program. The scalable parameter was calculated by comparing the volume for each knee element relative to the volume of the first knee soft tissue model (66 years old cadaver knee). As the CT scans covered a larger range of the legs than the MRI scans, the length of the CT-based bone models was edited first to have the same length for the CT-based and MRI-based models. The length adjustments were processed using Mimics program.
Table 1. Patient-specific 3D knee joint geometrical models and demographics.

<table>
<thead>
<tr>
<th>Model</th>
<th>Anterior</th>
<th>Posterior</th>
<th>Lateral</th>
<th>Medial</th>
</tr>
</thead>
<tbody>
<tr>
<td>C496R</td>
<td><img src="image1.png" alt="Image" /></td>
<td><img src="image2.png" alt="Image" /></td>
<td><img src="image3.png" alt="Image" /></td>
<td><img src="image4.png" alt="Image" /></td>
</tr>
<tr>
<td>66 yrs; Male; Right</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>C545R</td>
<td><img src="image5.png" alt="Image" /></td>
<td><img src="image6.png" alt="Image" /></td>
<td><img src="image7.png" alt="Image" /></td>
<td><img src="image8.png" alt="Image" /></td>
</tr>
<tr>
<td>72 yrs; Male; Right</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>C555R</td>
<td><img src="image9.png" alt="Image" /></td>
<td><img src="image10.png" alt="Image" /></td>
<td><img src="image11.png" alt="Image" /></td>
<td><img src="image12.png" alt="Image" /></td>
</tr>
<tr>
<td>80 yrs; Male; Right</td>
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<td></td>
<td></td>
</tr>
<tr>
<td>C580B</td>
<td><img src="image13.png" alt="Image" /></td>
<td><img src="image14.png" alt="Image" /></td>
<td><img src="image15.png" alt="Image" /></td>
<td><img src="image16.png" alt="Image" /></td>
</tr>
<tr>
<td>86 yrs; Female; Right</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>C590B</td>
<td><img src="image17.png" alt="Image" /></td>
<td><img src="image18.png" alt="Image" /></td>
<td><img src="image19.png" alt="Image" /></td>
<td><img src="image20.png" alt="Image" /></td>
</tr>
<tr>
<td>78 yrs; Female; Right</td>
<td></td>
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</tr>
</tbody>
</table>

Figure 4. Five MRI-based 3D knee models, consisting of 4 bone elements and 9 soft tissue elements, shown in four different views (anterior, posterior, lateral and medial).
CT-based knee models

The reconstruction of the knee bones models which were from the CT scan images were performed only for one cadaver (right leg; male; 66 years old). The procedure to create the 3D knee bones model from the CT scan images was the same as described with the MRI scan. Because of the big quantity of dicom files of the CT scan images, however, the files were divided into two different project files of the CT scan images to overcome the low speed computer processing. The CT scan images in the first project file consisted of the femur and patella (Figure 5(a)), and in the second project file consisted of the tibia and fibula (Figure 5(b)).

![CT scan images of the (a) femur and patella part and (b) the tibia and fibula.](image-url)
To get the bone elements mask, we also utilized the thresholding method for CT bone elements. The HU scale range of the default CT bone thresholding was between 226 and 3071. We generated four different color masks for femoral bone, patella bone, tibial bone and fibula bone as shown in the Figure 6(a) and 6(b) by using the region growing tool.

Figure 6. Segmentation of (a) the femur (green) and patella (yellow) and (b) the tibia (blue) and fibula (red).
Chapter 2

The masks quality of each bone element was also improved by manual editing of the masks. We also calculated each STL representation with the default medium quality setting to create four different 3D surface models of the bones (Figure 7). The same procedure and settings as described for the MRI segmentation were used to smooth the mesh models.

Figure 7. The CT-based 3D surface mesh of the femur and patella (left) and the tibia and fibula (right) before smoothing.

All knee bone models after smoothing were imported as STL files. Figure 8 shows all knee bones in the same coordinate system.

Figure 8. Triangular surface mesh (STL) of the knee after smoothing.
2.3. Registration of MRI-based and CT-based knee models

The reconstructed MRI-based and CT-based 3D knee models on STL format were loaded into the MRI project file in the Mimics program as shown in the Figure 9. Due to a different coordinate system between the MRI-based and CT-based knee models, we registered the CT-based knee bone models to the MRI-based models by using bones surface-based registration method available in Mimics. By plotting the CT-based model into the MRI based model coordinate system, we developed a new 3D model which was combining the bone structures of the CT-based model and the soft tissues structures of the MRI-based model. Furthermore, we could calculate a local anatomical coordinate system for the MRI-based model based on the anatomy of the knee bones from the CT-based model.

![Figure 9. The reconstructed MRI-based and CT-based 3D knee models before registration.](image)

The surface based registration was completed one by one for each bone based on the MRI-based and CT-based surfaces, using the STL-registration function in the Mimics program. Due to the limitation of the registration function, before executing the surface based registration, the CT-bones model position was translated to a position relatively close to the MRI-based model (Figure 10 (a)). The registration was then performed by setting the registration function into a global registration with a minimal point distance filter of 0 mm. The program automatically created the matrix transformation coordinate system for each bone registration procedure.
To assess the accuracy of the registration procedure each bone model registration, we collected the residual error (least square distance) value which was provided by the program. The registered CT-based models relative to MRI-based model are shown in the Figure 10 (b).

Figure 10. The MRI-based and CT-based knee models: (a) translated position of the CT-based model before registration; (b) CT-based model registered on the MRI-based model.

Figure 11. The four planes of view of the new 3D knee model, combining the bone models from the CT scans and the soft tissue models based on the MRI scans.
The new 3D knee geometrical model, combining all elements including CT-based bones elements and MRI-based soft tissues elements, is shown in Figure 11. This new model was used to develop a detailed anatomical model of the knee in The Anybody Modelling System (Dijkstra et al., 2012).

To assess the quality of the reconstruction process of bone structures in the same subject (cadaver C496R) from MRI and CT scan images, we compared the volume of the bone models, including femur, tibia, fibula and patella bones. Before calculating the geometrical differences, the bones of the MRI based CT based models were edited in Mimics to obtain the same length. Figure 12 shows the edited bone structures of the CT based and MRI based bone models. We calculated the normalized volume differences of the CT bone structures relative to the MRI bone structures.

![CT-based bone model relative to MRI-based bone model](image)

**Figure 12.** The registered CT-based bone model (blue) relative to the MRI-based bone model (red).

### 3. Results

The 3D knee geometrical models of five different cadaver knee were developed successfully from MRI and CT scan images as shown in the Figure 4 and Figure 8, respectively, as well as one knee model of cadaver C496R which combining the soft tissue models from MRI scan image and bone models from CT scan image (Figure 11). We only failed to develop the structure of the patellar cartilage from cadaver C580B as the image quality was very poor in this region.

The volumes of all elements of the MRI based knee models are listed in Table 1. The comparison of the element volumes of the four knee models relative to the first knee model (C496R) is plotted in Figure 13 (a-c). The volume comparison was divided into three categories i.e. bones, ligaments, and meniscus and articular cartilage. From these three categories (Table 1), in average, we found that model C555R had the biggest structures.

In general, the knees were not scalable in volume. However, for the bone structures, we found a better agreement of scalable volume than for the soft tissues structures. The maximum volume ratio of the bone structures was 11.9% for model C555R (Table 1). For the soft tissue structures, the average range variation of volume ratio ranged from 30.3% to 73.2% for the ligaments and from 23.6% to 59.9% for the cartilage and menisci. The standard deviations were quite high indicating that these volumes were not scalable.
Table 1. The volume of the knee joint components of the five models, and the volumes of each knee element relative to knee model C496R.

<table>
<thead>
<tr>
<th>Knee elements</th>
<th>Volume [mm³]</th>
<th>Ratio Volume relative to C496R volume [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>C496R</td>
<td>C545R</td>
</tr>
<tr>
<td>Femur</td>
<td>201063.5</td>
<td>233736.6</td>
</tr>
<tr>
<td>Tibia</td>
<td>167843.9</td>
<td>175943.4</td>
</tr>
<tr>
<td>Fibula</td>
<td>16026.4</td>
<td>16799.1</td>
</tr>
<tr>
<td>Patella</td>
<td>20797.0</td>
<td>23198.4</td>
</tr>
<tr>
<td><strong>Mean</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>SD</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ACL</td>
<td>1197.1</td>
<td>1491.1</td>
</tr>
<tr>
<td>PCL</td>
<td>2661.4</td>
<td>4087.8</td>
</tr>
<tr>
<td>LCL</td>
<td>2274.1</td>
<td>1987.1</td>
</tr>
<tr>
<td>MCL</td>
<td>2542.3</td>
<td>2407.5</td>
</tr>
<tr>
<td><strong>Mean</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>SD</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cartilage Patella</td>
<td>9691.5</td>
<td>5262.2</td>
</tr>
<tr>
<td>Cartilage Tibia</td>
<td>7283.7</td>
<td>4214.6</td>
</tr>
<tr>
<td>Cartilage Femur</td>
<td>19298.1</td>
<td>13039.7</td>
</tr>
<tr>
<td>Meniscus-Lateral</td>
<td>2821.5</td>
<td>3135.5</td>
</tr>
<tr>
<td>Meniscus-Medial</td>
<td>3910.9</td>
<td>3099.6</td>
</tr>
<tr>
<td><strong>Mean</strong></td>
<td></td>
<td></td>
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<tr>
<td><strong>SD</strong></td>
<td></td>
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</table>

Figure 13. Comparison of the volumes of the bones (a), ligaments (b), cartilages and meniscus (c) relative to the reference cadaver (C496R: male, 66 years).

The registration error of the femoral bone, tibial bone, fibula bone and patella bone were 1.02 mm, 0.91 mm, 0.44 mm and 2.05 mm, respectively. The relative geometrical differences of the CT based model relative to the MRI based model are shown in Table 2. The volume of the
CT based models were bigger than the MRI based model with an average difference of 12.5% (SD= 15.4%). The smallest different volume was the fibula bone of 2.1% and the biggest different volume model was the patella bone of 35.4%. The numbers of triangles and points were similar. From four bones, we found that the numbers of the triangles and points of the CT based tibial bone were smaller than the MRI based tibial bone.

Table 2. The volume, number of triangles and points of all knee elements between the MRI-based and CT-based 3D knee models.

<table>
<thead>
<tr>
<th>Knee elements</th>
<th>MRI-(Red)</th>
<th>CT-(Blue)</th>
<th>Relative Difference [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Femora</td>
<td>207233.5</td>
<td>54610.0</td>
<td>27307.0</td>
</tr>
<tr>
<td>Tibia</td>
<td>168800.6</td>
<td>59550.0</td>
<td>29779.0</td>
</tr>
<tr>
<td>Fibula</td>
<td>15976.7</td>
<td>12410.0</td>
<td>6207.0</td>
</tr>
<tr>
<td>Patella</td>
<td>18496.1</td>
<td>13340.0</td>
<td>6672.0</td>
</tr>
<tr>
<td><strong>Mean</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>SD</strong></td>
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</table>

4. Discussion

The development of a subject specific 3D FE model of the knee joint can offer a solution to optimize the ACL reconstruction results. Using such a model, the knee mechanics and kinematics can be optimized prior to surgery, providing a surgical plan for the placement of the ACL graft. The geometrical model is an important aspect to enable the generation of a reliable FE simulation. Therefore, we saw the need to describe some parameters which are important to consider during model development. Many parameters influenced the development of the model. In the current study we described the comparison of the model reconstruction process between CT-based and MRI-based bone models as well as the quantification of the scalable factor amongst the volumes of the various knee soft tissues from the different knees.

In general, the volumes of the three large bones (tibia, femur and fibula) based on the CT data were comparable to the MRI-based models with the maximum volume difference being 6.6%. The result showed that the bone segmentation from MRI images was relatively good. We also succeeded to develop a new knee subject specific model (Figure 11) of cadaver C496R by combining the two different images modalities i.e. MRI based soft tissue models and CT based bone models. Relative to minimum slice thickness image scan of 0.5 mm for MRI images and 0.6 mm for CT images, we considered that the maximum error of 1.02 mm during the registration process for the three long bones (tibia, femur and fibula) was still acceptable.

In our current study, we still used CT scan and MRI scan to develop the knee geometrical model. By using CT scan images with an intact leg, some possibilities to develop a more accurate 3D knee geometrical model can be realized. The possibilities are to calculate bone material properties based on gray scale value combined with a bone phantom calibration (Keyak and Falkinstein, 2003) as well as to quantify a knee joint coordinate system which includes the femoral head (proximal part of the femur) and the malleolus (distal part of the tibia) (Grood and Suntay, 1983; Wu et al., 2002). Hence, for some applications it can be useful to combine the two scanning modalities, but this will depend on the application of the models and on the ethical
issues related to the double scanning procedure. Although CT scans require a shorter scanning time than MRI, the scanning involves exposure to radiation. Furthermore, by using two types of images, extra time is required, costs (financial) are higher as well as extra data processing including segmentation and registration process.

We believe that, relative to CT, using MRI scans is more effective for ACL research because we can adequately reconstruct the bony segments (Rathnayaka et al., 2012) and the soft tissues (ligaments, cartilages and meniscus). Furthermore, MRI is non-invasive and therefore safer than CT. Obviously, MRI does not allow for representation of the bone material properties in the different bony segments and therefore will have to be simulated as rigid body structures as researchers have done in the past (Besier et al., 2005; Zhixiu Hao et al., 2007; Netravali et al., 2011; Donahue et al., 2003; Potocnik et al., 2008). In clinical practice, MRI is a common diagnostic tool and advised to perform before undertaking an arthroscopic procedure.

Most of the MRI images do not represent a full bone of femur, tibia and fibula. This makes it difficult to define the anatomical-based coordinate systems. However, it is still possible to define the anatomical coordinate system of such short images by applying the method which is recommended by Miranda and colleagues, who determined an anatomical coordinate system for 3D bone models of isolated human knees (Miranda et al., 2010).

In the present study defining the anatomical structures in detail to develop the 3-D knee models was a very time consuming process. The difficulties were not only defining the soft tissues, but also the attachment sites of ligaments and the boundaries of some bony parts were also difficult to detect. In bony parts, it was difficult to determine the cortex region of the bone and separate that from the cartilage layer. This possibly was also one of factors which influenced the big difference of volume of the patella bone (35.4%) between CT and MRI, as well as the bigger volume of the CT based bone models. The thresholding method during the reconstruction process was very difficult to handle as it was trial-and-error to find an appropriate setting for covering most of bone surfaces in the MRI images. On the other hand, by this method the segmenting of the bony surfaces by CT imaging was fast. The fact that performing the manual segmentations was not always easy is related to the quality of scan images, a lack of anatomical knowledge, segmentation skills (Lee et al., 2008) and the cadaver condition.

An obvious limitation in this study was the use of relatively old knee donors (average 76 years old), whereas ACL-patients are typically quite young. Younger donors were not available. We used the clinical MRI scanning protocol which is applied for living knee patients. There are obviously other MRI-protocols (with much longer scanning time) which can be used for cadaver materials and which would probably give a better quality image of the knee structures. As suggested by the Moro-oka study (Moro-oka et al., 2006), MRI images can be improved to identify both bone and soft tissue structures by using different types of MRI sequences. Lower magnetic fields (0.3 T) will give a good resolution for bone segmentation, whereas higher magnetic fields increase signal intensity for better soft tissue resolution. The downside of this approach is obviously the increase in scanning capacity, time and costs. In any case, we chose for a rather simple, clinically used MRI protocol because the goal of the study is to apply our techniques to living patients and we do not want to deviate too much from current clinical practice.
To reduce the segmentation time for individual knees, we also evaluated the scalability of knee geometries in this study. We found that there were no scalable factors of the geometrical model. The volumes of each of the knee structures varied amongst the models, especially for the soft tissue structures. These differences were not only caused by the fact that every knee is unique (and not simply scalable), but we also realize that the method to evaluate the scalable factor which is only based on volume may not have been accurate and strong enough. The measured volume of the knee structures was very sensitive to adding and removing some pixels during segmentation. This change was even larger in the soft tissue structures which have smaller total volumes than bone structures. For example when segmenting one extra slice in the sagittal view of the LCL and MCL, this will give a large change in the thickness of these structures and this will considerably influence the total volume of the structures. This effect is less in larger volume structures such as bone. This explains that we found better volume ratios for long bone structures such as tibia, femur and fibula. To speed up the segmentation process, perhaps a semi or integrated automated segmentation of the soft tissues such as segmentation of bone cartilages and patella bone (Baldwin et al., 2010) and lateral meniscus could be developed. (Swanson et al., 2010).

5. Conclusions

By developing a subject-specific 3D knee geometrical model, this study aimed to compare the bone model reconstruction quality between CT and MRI scan images. Furthermore, an assessment was made of MRI based segmentations of the soft tissues and whether these are scalable in terms of segment volumes. The results showed that the MRI based bone model had a geometrical size comparable with the CT based bone model. Unfortunately, we could not find a good scalable factor amongst the knee element structures. These results indicate that MRI can be used as a single image source to develop a FE 3D knee geometrical model especially to analyze patients who require an ACL reconstruction. Further development of a faster segmentation tool is necessary for broad clinical application.

References

Chapter 2


