Accuracy of 3D Cartilage Models Generated From MR Images Is Dependent on Cartilage Thickness: Laser Scanner Based Validation of In Vivo Cartilage

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Abstract

Cartilage morphology change is an important biomarker for the progression of osteoarthritis. The purpose of this study was to assess the accuracy of in vivo cartilage thickness measurements from MR image-based 3D cartilage models using a laser scanning method and to test if the accuracy changes with cartilage thickness. Three-dimensional tibial cartilage models were created from MR images (in-plane resolution of 0.55 mm and thickness of 1.5 mm) of osteoarthritic knees of ten patients prior to total knee replacement surgery using a semi-automated B-spline segmentation algorithm. Following surgery, the resected tibial plateaus were laser scanned and made into 3D models. The MR image and laser-scan based models were registered to each other using a shape matching technique. The thicknesses were compared point wise for the overall surface. The linear mixed-effects model was used for statistical test. On average, taking account of individual variations, the thickness measurements in MRI were overestimated in thinner (<2.5 mm) regions. The cartilage thicker than 2.5 mm was accurately predicted in MRI, though the thick cartilage in the central regions was underestimated. The accuracy of thickness measurements in the MRI-derived cartilage models systemically varied according to native cartilage thickness.

1 Introduction

A substantial portion of the elderly population suffers from degenerative joint disease (osteoarthritis). However, the primary treatments address the symptoms rather than the disease since the causes for initiation and progression of osteoarthritis are not well understood [1]. Quantitative measurement of cartilage thickness offers an attractive method for learning more about the disease and can be used to track the progression of osteoarthritis [2]. In particular, semi-automatic image segmentation methods have been used to delineate cartilage boundaries of knee MR images and to create 3D cartilage models [3–6]. While these 3D models have advanced our understanding of morphological cartilage changes associated with osteoarthritis [2,7–10], the evaluation of new disease modifying interventions would benefit from a rigorous assessment of the condition influencing the spatial accuracy of these models.

Assessing the regional accuracy of cartilage morphometric measurements using 3D models derived from MRI is particularly important for the evaluation of disease modifying interventions designed for treating osteoarthritis at an early stage since the lesion is typically localized to a specific region of the cartilage. Thus more global measurements such as cartilage volume might not detect regional changes that are critical to the assessment of the disease process. Thus there is a need for additional work to complement existing studies (Table 1) of 3D cartilage models; that have assessed the accuracy of volume measurement of a 3D cartilage model using a water displacement method for cartilage in specimens from amputated knees and total knee replacement (TKR) surgeries [3,8,10,11]; and surface area of a 3D cartilage model using an image of aluminum foil covering the cartilage from total TKR surgery [10]. An accurate stereophotogrammetric method has also been used to make precise measurements of the 3D shape of ex vivo cartilage [12] and to verify cartilage thickness measurements. However, that study was conducted using cadaveric cartilage from MR images [13] and the relationship of these measurements taken from in vivo MRI has not been established. While thickness accuracy has also been evaluated using anatomical sections [8,10,11], it was unclear whether these studies controlled for thickness changes to tissue dehydration in open air.

Thus while many aspects of the accuracy of morphometric have been addressed some of the conditions limiting the accuracy of morphometric measurements derived from MRI including partial volume effects, chemical shift, and geometric distortion [14] have not been evaluated. The partial volume effect is particularly important when considering the sensitivity of the accuracy of thickness measurements to the native cartilage thickness. The accuracy was predicted to be reduced for thinner cartilage due to a partial volume effect and MR image voxel anisotropy [15,16].

There remain unanswered questions regarding the sensitivity of cartilage thickness measurements to the native thickness of the cartilage. Considering the importance of local regional cartilage thickness measurement in diagnosing the progression of osteoarthritis, there remains a need to evaluate the sensitivity of cartilage thickness measurements to the native thickness of the cartilage. Thus, the purpose of this study was to assess the accuracy of in vivo cartilage thickness measurements from MR image-based 3D cartilage models using a laser scanning method [17,18] and to test the hypothesis that the accuracy of the thickness measurement changes with cartilage thickness.

2 Methods

2.1 Materials

Approval to conduct this study was obtained from the institutional review board. Total knee replacement patients who were scheduled for surgery within 1 month were recruited. Subjects with reasonable amount of cartilage in at least one compartment in the osteoarthritic knee were included. After informed consent was obtained, MR images of the osteoarthritic knee were performed on ten subjects (all male, age 69±8 years, height 173±6 cm, weight 99±17 kg, and body mass index 33±5 kg/m^2). To minimize the effect of loading history on articular cartilage morphology, the MR images were taken early in the morning and the subjects were instructed to avoid loading knees before MR imaging. These subjects were to be operated on within 2 weeks (an average of 6 days) after the MR images were obtained.

2.2 MR Imaging Knee Cartilage

Knee MR images were obtained prior to surgery using a fat-saturated 3D spoiled gradient recalled echo in sagittal plane. This sequence had TR (repetition time) =60 ms, TE (echo time) =5 ms, flip angle =40 deg, receiver bandwidth of +/−15.63 kHz, FOV (field of view) of 140×140 mm, slice thickness of 1.5 mm, 60 slices, and matrix of 256×256. So the in-plane resolution was 0.55×0.55 mm. Scan time was 10 min and 18 s. Images were acquired on a 1.5T MRI scanner (GE Healthcare, Milwaukee, WI and SIGNA LX, software rev. 12.4) with a standard transmit-receive extremity coil. This sequence has previously been used to assess cartilage morphology [6,8,19,20].

2.3 Cartilage Model From MR Images

The MR images were processed using custom software [6]. A single experienced observer performed all segmentations to achieve maximum reproducibility [6]. The cartilage was segmented from each image of a set of MR images using a semi-automatic B-spline snake method [5,21]. Boundaries from the initial segmentation were manually corrected to increase accuracy. The boundaries obtained from the segmentation were reconstructed into a 3D polygon model using a surface rendering method [22]. The reproducibility of this method was tested in our previous study [6].

2.4 Cartilage Model From Actual Tissue Using Laser Scan

Following total knee replacement surgery, the resected tibial plateaus were kept in isotonic saline solution for less than 2 h until they were laser scanned. The actual cartilage thickness was measured using a 3D laser-scan based method. The proximal tibia was attached to a platform made of acrylonitrile butadiene styrene (ABS) using instant epoxy glue and left in air for 10 min for adhesive fixation. The cubical ABS platform was designed with two cylindrical bars to exactly register multiple laser-scan data in order to reconstruct the 3D shape. The fluid on the cartilage surface was dried in air during the adhesive fixation time for 10 min and the shiny and semitransparent surface was treated with white spray powder (starch/acrylates/acrylamide copolymer) to increase opacity and decrease the laser-scan errors [16].

The cartilage surface was scanned on a rotating table using a 3D desktop laser scanner (Model-15, Cyberware, Monterey, CA). The resolution of the laser scanner was 50 μm in depth. The field of view of the laser scanning was 80×80 mm^2 and points were sampled every 300 μm. The scanning was repeated at 24 different angles and registered using the program CYDIR (Cyberware, Monterey, CA). This overlapping multiscan technique increased the overall accuracy to 50 mm. The laser scan to obtain a 3D surface geometry took 12 min. After the articular surface of the cartilage had been scanned, the proximal tibia

piece was submerged in 6.0% sodium hypochlorite solution for up to 24 h to digest the cartilage [6]. After the cartilage was removed, the laser scanner was again used to find the subchondral bone. The two 3D laser scans of cartilage and subchondral bone surfaces were registered to each other using the features in the ABS platform by rigid body rotation and translation [23] (Fig. 1). The registered surfaces were reconstructed into a 3D model of actual cartilage (Fig. 1) using the RAPIDFORM software (Inus Technology, Seoul, South Korea) [6].

2.5 Thickness Comparison Between Two Models

Thicknesses were calculated for all the points on the articular surface by finding the closest point on the cartilage-subchondral bone interface surface. The thickness information was encoded on the cartilage model [5,6] as a thickness map. The 3D cartilage models from laser-scan data and MR images were registered to each other using a surface matching technique [23]. The thickness maps were projected onto planes and compared point wise as shown in Fig. 2. The difference map was calculated by subtracting thickness values of the laser-scan based 3D model from the MR image-based 3D model.

2.6 Linear Mixed-Effects Model

We hypothesized that the errors of the thickness measurements varied linearly with actual cartilage thicknesses. The statistical model to test this hypothesis included the effects of not only actual thicknesses but also variations between specimens on the errors because the data were obtained from ten specimens [24].

$$y_{ij} = (\beta_0 + b_{0i}) + (\beta_1 + b_{1i})x_{ij} + \varepsilon_{ij}$$

In the formula, y_{ij} is the error of jth measurement in ith specimen and x_{ij} is its actual thickness. β_0 and β_1 represent the fixed-effects interpreted as population parameters for the intercept and the slope, respectively. b_{0i} and b_{1i} represent the random-effects, which are deviations from population parameter of ith specimen for the intercept and the slope, respectively. ε_{ij} is the residual. The null hypothesis that $\beta_1 = 0$ was tested which means that the error is not the linear function of actual thickness. The 95% confidence interval of the fitted line was also calculated to test if error is different from 0.

2.7 Error Assessment in Measuring Actual Cartilage Shape

The sensitivity of thickness change due to dehydration was confirmed in a manner similar to a previous study that reported the dehydration rate of articular cartilage in the open air and provided an approximation function of cartilage thickness reduction over time [25]. We confirmed that the previously reported approximation function for the type of specimen in this study by evaluating the volume change due to dehydration over a 24 h period. The results were consistent with the previous study. The time the cartilage specimens were exposed to air were carefully controlled to 10 min of preparation and 12 min of laser scanning. Using the approximation function the thickness change for 22 min of exposure to air was less than 1%.

3 Results

The cartilage thickness based on 3D cartilage models derived from MRI tends to be overestimated for thin cartilage and the error decreased with the increase in the cartilage thickness for all specimens (Fig. 3).

The accuracy systematically varied with the actual cartilage thickness at the significance level of 0.05 according to the linear mixed-effects model. The fitted line along with 95% confidence interval for the fixed-effects was obtained from a linear mixed-effects model (Fig. 4). The slope and intercept from the linear mixed-effect model was −0.25 and 0.71, respectively. The variations of slope and intercept between individual specimens ranged from −0.44 to −0.07, and from 0.23 to 1.00, respectively. The residuals of the mixed effects models were evenly distributed along the fitted values in a visual check.

On average, taking account of individual variations, the thickness measurements in MRI were overestimated in thinner (< 2.5 mm) regions. The differences approached zero at the thickness between 2.5 mm and 3.3 mm, where the 95% confidence interval included the zero line. In thicker (>3.3 mm) regions the thicknesses were underestimated.

4 Discussion

The results showed that the accuracy of cartilage thickness measurements in the cartilage models derived from MRI varies systematically. Thin cartilage tends to be overestimated in the cartilage models. The thickness measurement was accurate for the cartilage whose thickness was between 2.5 mm and 3.3 mm.

The linear mixed-effects model could explain the effects of actual thicknesses (fixed effect) and specimens (random effect) on the errors in measuring cartilage thickness in MR image-based 3D models. A significant linear relationship between the accuracy and the actual cartilage thickness was found. This result supports the previous simulation studies [15,16] that the thickness of thin objects such as articular cartilage is overestimated due to partial volume effect and MR imaging voxel anisotropy.

The study showed that the accuracy of thickness measurement of articular cartilage in MRI-derived cartilage models is affected by its actual thickness. In our MR images, tibial cartilage orientation was almost vertical to sagittal plane thus the orientation effect was negligible but the MR image voxel anisotropy was 2.7 (=slice thickness/in-plane resolution), which is in the range that thin cartilage less than 2.5 mm can be affected [15,16].

This study implies that the accuracy of cartilage thickness measurements in cartilage models should be considered in the context of the native thickness of the articular cartilage specimens. It is expected that the error would be lower in measuring cartilage thickness of healthy articular cartilage.

While the previous simulation studies showed that the error approached zero in thick cartilage [15,16], the linear mixed-effects model in this study suggested that the thick cartilage (>3.3 mm) can be underestimated. One of the possible explanations should be due to our assumption of the linear change in the accuracy and testing thin osteoarthritic cartilage specimens, which may have governed the slope and the intercept of the linear mixed-effects model. While weight bearing was minimized prior to testing it is possible that the thickness in the central regions of cartilage was underestimated due to unknown residual compressive deformation in the central regions of the articular cartilage resulting from weight bearing prior to the test or passive joint capsule tension [26,27].

The specimens were from osteoarthritic knees, so some portions of the specimens had surface fibrillation. Though it did not change the surface shape visibly and was not detectable in MRI, we did not quantify the effect of this surface fibrillation on the laser scanning technique.

A thickness change due to dehydration of the cartilage in open air is an important consideration in studies of this type. The specimens were carefully controlled to be in the open air for 22 min from the starting of the set up to the completion of the laser scanning, 10 min for fixation on the platform and 12 min for scanning. While there could have been a small amount of dehydration during the 22 min there was no observable change in the shape and based on a study by Pham et al. [25] the cartilage thickness reduction over 22 min was less than 1%. Thus while small this effect should be considered when evaluating the results presented here or in other studies of this type.

We tested the accuracy of only one MR protocol and one resolution. There are many variables in MR imaging such as sequence, image resolution, and magnet strength. Recently, more efficient protocols such as steady state free precession (SSFP) [28] and fluctuating equilibrium MR imaging [29] have been developed for cartilage imaging and 3.0 Tesla machines are being adapted in clinics. This would save imaging time, increase image resolution, and signal to noise ratio (SNR). Currently, 3.0 Tesla magnets can achieve higher resolution such as in-plane resolution of 0.31 mm or less [30,31]. For the purpose of this study we assumed that the semi-automated B-spline snake method would give the best segmentation results. However, there exist other segmentation schemes such as anatomical template deformation or more automated segmentation techniques [32]. It may be that for other MR protocols and segmentation methods, the accuracy of cartilage thickness measurement will be different than what we found in this study. Thus, additional studies with different settings in the future would help generalize this result.

In conclusion, the accuracy of cartilage thickness measurement from MRI-derived cartilage models varied according to cartilage thickness. The thickness of thin cartilage less than 2.5 mm was significantly overestimated in MRI, with MRI in-plane resolution of 0.55 mm and thickness of 1.5 mm when processed using semiautomated B-spline snake method. The cartilage thicker than 2.5 mm was accurately estimated in MRI though the thick cartilage in the central regions was underestimated.

Acknowledgments

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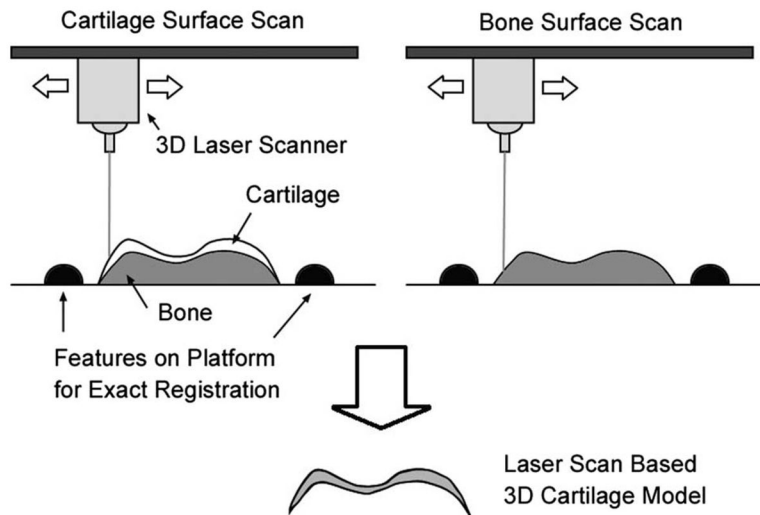
Fig. 1.
The schematic description of creating 3D cartilage models from two laser-scan data sets. Cartilage surface and bone surface were scanned along with features on platform and registered to each other using the platform features to obtain actual shape of cartilage.
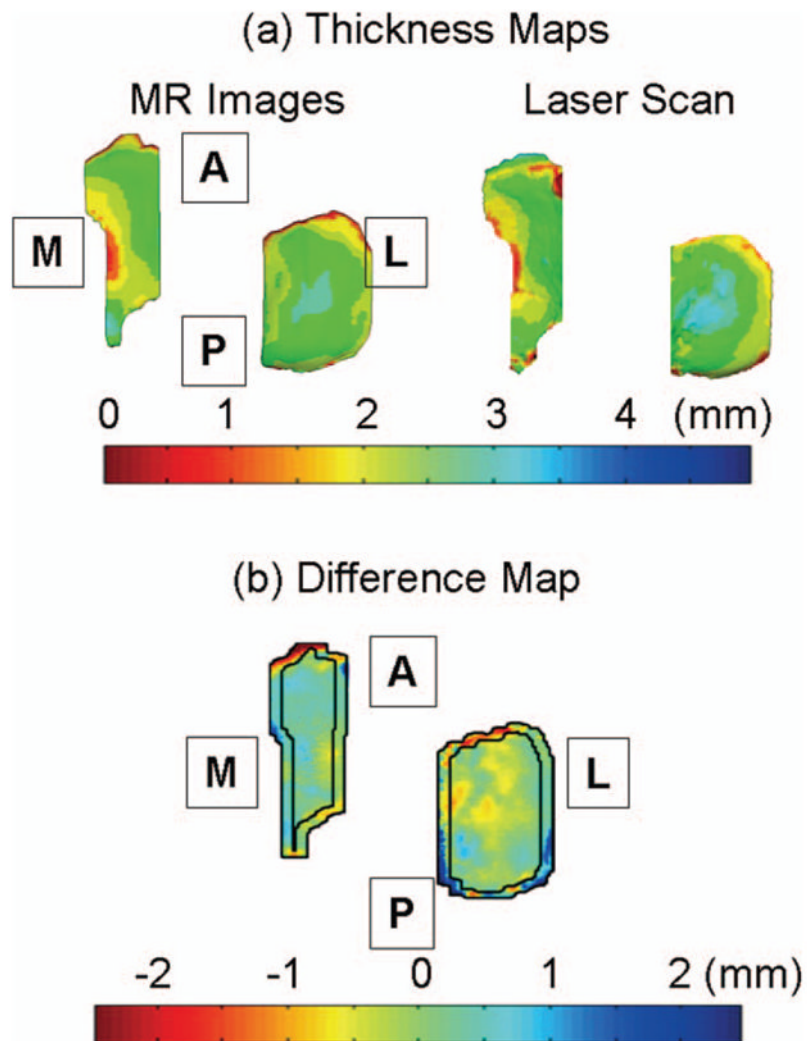
Fig. 2.
(*a*) Thickness maps calculated from a MR image-based 3D model and a laser-scan based 3D models. Dark blue represents the thickest cartilage and dark red represents the thinnest cartilage as shown in the color bar. (*b*) Difference maps were calculated by subtracting the thickness maps from the laser-scan based 3D model from the MR image-based 3D model. Thus blue regions represent thickness overestimation in MR image-based 3D models, green regions show the same estimation, and red regions represent underestimation. The letters A, P, M, and L in boxes represent anterior, posterior, medial, and lateral, respectively.
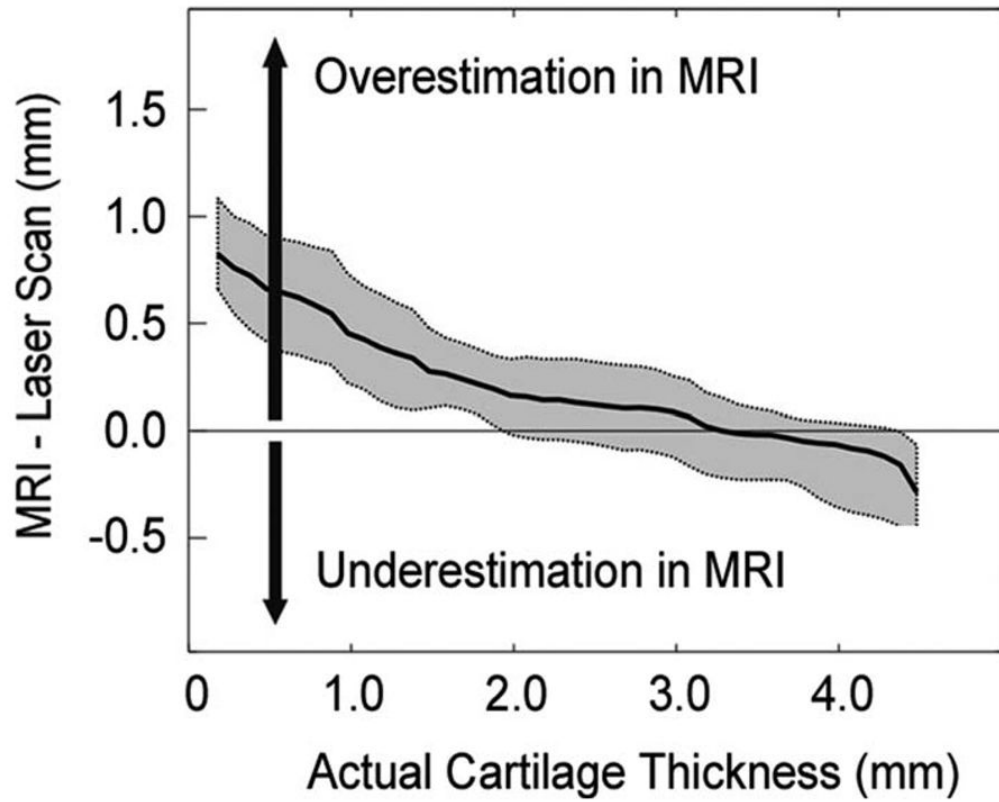
Fig. 3.
Differences in thickness measurement between MRI based 3D models and laser-scan based 3D models were calculated point by point for each of the specimens. This graph is from one of the tested specimens. Positive values represent thickness overestimation in MR image-based measurements. The median of the difference was drawn as solid lines, and 25% and 75% quartiles were drawn as dotted lines.
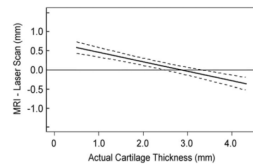
Fig. 4.
Variations of individual specimens are shown for ten specimens with gray lines. Positive and negative values represent thickness overestimation and underestimation in MR image-based 3D models, respectively. The linear mixed-effects model was used to find a fitting line of the data from ten specimens. The black lines represent the fitted line from the linear mixed-effects model and the 95% confidence interval.

Table 1

Summary of previous studies on the accuracy of MRI-derived 3D cartilage models

Validation method	Specimens	No. of specimens	Error quantification
Cartilage volume	Femoral, tibial, and patellar cartilage from amputated knees, and patellar cartilage from TKR surgery [3] Tibial cartilage from TKR surgery [8] Tibial and patellar cartilage from TKR surgery [10] Femoral, tibial, and patellar cartilage from cadaveric human knees [12]	3+3 8 Half of 21 8	8.2% and 5.9% for two MRI methods rms average 5.5% qMRI overestimation in patella (5.1%) and lateral tibia (3.6%), and underestimation in medial tibia (−3.1%) 3.6%, 3.8%, 4.2%, and 1.1% for patellar, femoral, medial, and lateral cartilage, respectively
Cartilage surface area	Tibial and patellar cartilage from TKR surgery [10]	21	4.6% to 9.1%
Cartilage thickness	Tibial cartilage from TKR surgery [8] Tibial and patellar cartilage from TKR surgery [10] Femoral, tibial, and patellar cartilage from cadaveric human knees [12]	8 Half of 21 8	rms average 6.2% 4.3% for patella and 12.3% for medial tibia For patella, 84.5% of the surface <1 mm error