The objective of this work was to develop a computational approach for quantifying the three-dimensional (3D) thickness distribution of articular cartilage with magnetic resonance (MR) imaging, independent of the imaging plane, and to test the reproducibility of the method in the living. An algorithm was implemented, based on a 3D Euclidean distance transformation, and its accuracy was assessed in geometric test objects, for which an analytic solution was available. The precision of the method was evaluated in six replicated MR data sets of the knee joint cartilage of eight volunteers. The algorithm produced 3D thickness values identical to those of the analytic solutions in the test objects. The reproducibility of the mean cartilage thickness in the patellar and tibial cartilages was 1.5–3.4% (root-mean-square average of the individual coefficient of variation percent), that of the maximal thickness 2.1–7.9%, and that of the thickness distribution 2.3–6.1%. The method presented allows for noninvasive analysis of 3D cartilage thickness from MR images in biomechanical and clinical investigations. Magn Reson Med 41:529–536, 1999. © 1999 Wiley-Liss, Inc.

Key words: cartilage thickness; MR imaging; reproducibility; 3D reconstruction

Accurate data on the quantitative distribution of articular cartilage in diarthrodial joints are required in clinical management as well as in fundamental research. They are useful for screening, staging, and monitoring joint disease, and for controlling the success of therapeutic interventions, such as chondroprotective medication, physiotherapy, correction osteotomy, or cartilage transplantation (1). In biomechanics, potential applications include the design of computer models of joint kinematics and load transmission (2,3), the study of functional adaptation of cartilage to mechanical loading (4), and the investigation of in situ cartilage deformation in intact joints, in both cadaver specimens (5,6) and living subjects (5,7).

It has been shown that MRI is capable of providing accurate data on cartilage volume (8–12) and thickness (11–14), if high-resolution T1-weighted, fat-suppressed, three-dimensional (3D) gradient-echo sequences are employed. Since in geometrically complex joint surfaces the appropriate distance vectors are not always located within the MR image plane, 3D techniques are required that take into account the spatial orientation of the thickness measurements within the cartilage layer (11,15–19). Such techniques make it possible to determine the cartilage thickness distribution throughout entire joint surfaces, independent of the respective section location and orientation, and this is of particular importance for obtaining a satisfactory reproducibility in longitudinal studies, in which identical section locations and orientation cannot be readily reproduced (20,21). In previous studies 3D cartilage thickness measurements (based on 3D reconstruction of MR images) have been presented (11,16–21), by demonstrating thickness maps throughout human knee joint surfaces. However, some of these techniques (11,16–18) have suffered from limitations:

1. Due to the discrete representation of the data, the normal vectors were only defined in a few discrete directions, giving rise to inaccurate measurements around cartilage lesions and surface edges.

2. Because each cartilage plate was defined as one object, surface-to-surface measurements were generated by forcing the normal vectors of opposite sides to be "almost" antiparallel to each other (16–18). This limitation may lead to artifacts at the edges of the cartilage plates, the vectors running obliquely through the layer. Although this did not interfere with the visualization of cartilage thickness data (because only the joint edges were affected), no statistically accurate computation of maximal and mean cartilage values and standard deviations of the thickness measurements could be provided. These parameters are, however, valuable for the precise quantification of cartilage thickness changes over time (20,21) or, for instance, during cartilage compression (6). In particular, they allow for improved statistical analysis in cross-sectional studies, when trying to identify individuals with abnormally high or low cartilage thickness.

In the present paper we aim to overcome these problems by developing a new computational technique, based on separate segmentation of the cartilage surface and the bone-cartilage interface and on 3D Euclidian distance transformation; this method avoids the explicit calculation of normal vectors. The specific objectives of this study were a) to propose a new algorithm for statistically accurate 3D determination of the mean and maximal cartilage thickness in articular surfaces (independent of the specific section position and orientation); b) to test the performance of this algorithm in geometrically defined objects; and c) to assess the reproducibility of MRI-based cartilage thickness measurements, by applying the algorithm to replicated...
MATERIALS AND METHODS

Surface Generation

The cartilage is segmented from the surrounding tissue on a slice by slice basis using a region-growing algorithm whereby the seed point is placed manually in the middle of the cartilage (11,16,17,20). The result is controlled visually and is manually corrected, if necessary. Furthermore, for the separation of the cartilage surface and the cartilage-bone interface, the transition points between both surfaces were marked on the cartilage boundary by the user in each section. As the in-plane resolution and the slice thickness differ, the segmented cartilage regions of each slice are interpolated into an isotropic binary volume, using a shape-based interpolation method (22,23) that estimates the boundary of the interpolated cartilage from the weighted distances to the cartilage boundaries in the two adjacent slices. A standard boundary tracking algorithm was then implemented to extract and separate the two cartilage surfaces from the interpolated cartilage regions, thus providing an isotropic binary volume image in which the voxels of the cartilage surface and the cartilage-bone intersection have different labels.

Computational Method for 3D Cartilage Thickness Measurements

An algorithm was designed to calculate the cartilage thickness, based on a 3D Euclidean distance transformation (EDT), which is an approved image transformation in the field of computer vision (24). It allows the accurate and efficient calculation of the 3D thickness distribution between two extracted 3D surfaces avoiding at the same time the difficulties in defining and computing normal vectors on the discrete surfaces. The EDT operates on isotropic binary volume data, whereby voxels belonging to segmented surfaces are labeled. The resulting 3D distance map assigns to each voxel of the volume the distance to the closest surface voxel. The algorithm that transforms the binary volume into a distance map is described in the three following sections.

Initialization

The voxels are represented by the index vector \( \vec{v} = (v_1, v_2, v_3) \) with 0 < \( v_i \leq N_i \), where \( N_i \) is the number of voxels in each direction of the 3D image data. In the distance map \( d(\vec{v}) \), the voxels belonging to the surface are initialized with zero, whereas all other voxels are set to infinity:

\[
d(\vec{v}) = \begin{cases} 
0 & \text{if } \vec{v} \in \text{surface} \\
\infty & \text{else} 
\end{cases} . \tag{1}
\]

Since the direction coordinates to the closest surface voxel have to be tracked during the EDT, an additional vector coordinate map \( \vec{u}(\vec{v}) \) is initialized:

\[
\vec{u}(\vec{v}) = [u_1(\vec{v}), u_2(\vec{v}), u_3(\vec{v})] = \begin{cases} 
(0, 0, 0) & \text{if } \vec{v} \in \text{surface} \\
(\infty, \infty, \infty) & \text{else} 
\end{cases} . \tag{2}
\]

Transformation

The principle of the EDT is the following: Local distance masks are defined, storing the displacement vectors between neighboring voxels. By passing these local distance masks over the initialized distance map, the global distance values are updated iteratively on the basis of the local distances, propagating them over the whole data volume. More precisely, the distance masks depicted in Fig. 1 contain direction vectors \( \vec{q}(l) \) pointing from the center of the mask to the neighboring voxels. While the masks pass across the volume, the new distance value \( d^{\text{new}}(\vec{v}) \) and its direction coordinates \( \vec{u}^{\text{new}}(\vec{v}) \) are updated at the considered voxel \( \vec{v} \) for each mask by the following assignment:

\[
\vec{u}^{\text{new}}(\vec{v}) \leftarrow \arg \min_{\vec{l} \in \text{mask}} \left( \| \vec{u}(\vec{v} + \vec{l}) + \vec{q}(l) \|_2 \right) \tag{3}
\]

and

\[
d^{\text{new}}(\vec{v}) = \| \vec{u}^{\text{new}}(\vec{v}) \| = \sqrt{u_1^2(\vec{v}) + u_2(\vec{v}) + u_3(\vec{v})} , \tag{4}
\]

where \( \| \) represents the Euclidean norm of the 3D Cartesian space and \( \vec{l} \) an indexing vector pointing from the center of the mask to its elements. This means that after placing the mask with its center at the considered voxel \( \vec{v} \) and adding the local direction vectors \( \vec{q} \) of the mask to the coordinate map \( \vec{u} \) at the corresponding positions \( \vec{l} \), one chooses from the resulting vectors the one with the smallest norm as new entry in the coordinate map.

To calculate the correct Euclidean distance from each volume voxel to the nearest surface voxel, different masks have to be moved across the data volume in a complex manner (Fig. 1), which can be roughly divided in a forward and backward pass. At each position encountered during the passes, the new direction coordinates are calculated according to Eq. [3]. In the forward mode, four passes are required in total for every slice, beginning at the bottom of the volume. In each slice, the mask F1 moves for each line, beginning at the top line, from left to right while mask F2 runs in the opposite direction from right to left (Fig. 1). Once arrived at the bottom line, masks F3 and F4 move in the opposite sense through the slice starting at the bottom line, whereas in each line F3 runs from right to left and F4 from left to right. The backward pass operates in a similar way, starting at the top of the volume (Fig. 1).

This updating process causes the local distances between neighboring voxels to propagate over the whole volume, every surface voxel acting as a source for a so-called distance wave. These wave fronts are expanded until they meet wave fronts coming from other surface voxels. Finally, a complete 3D distance map of the surface is constructed, encoding in each voxel the minimal distance measured normal to the nearest surface voxel (Fig. 2b).
3D Distance Assignment

To determine the distance between the cartilage surface and the bone-cartilage interface, mathematically two definitions are possible. First, the distance normal to the cartilage surface, i.e., the length of the normal vector originating on the cartilage surface and ending on the bone-cartilage interface, and second, the distance normal to the bone-cartilage interface. The latter was selected here (see Fig. 4c,d), because Lösch et al (16) have shown that this method is superior for the delineation of focal cartilage lesions. Therefore the EDT was applied to the bone-cartilage interface, yielding a complete 3D distance map of this surface. The distance values were then read out at those positions in the distance map that correspond to the voxels of the cartilage surface. This is mathematically equivalent to measuring the distance between both surfaces normal to the cartilage-bone interface (Fig. 2b,c). In conclusion, the EDT provides the exact minimal distance between the cartilage surface and the cartilage-bone interface; it is normal to the cartilage-bone interface but avoids the explicit calculation of normal vectors.

Test Models for the 3D Euclidian Distance Algorithm

We chose two 3D test models for the 3D EDT algorithm, an elliptical cylinder and a sinusoidal modulated ramp, for which the theoretical distance distributions could be computed analytically (Fig. 3). The analytical calculations were compared with the mean and maximal thickness obtained by the algorithm. A second measure of accuracy was obtained by testing the reproducibility of the method with respect to rotations and translations of the test models in the 3D space. Thus the error introduced by the interpolation of the voxels into a discrete lattice is determined. Each model was therefore randomly oriented and translated 10 times, and the 3D EDT algorithm was applied to the models in each of these positions. Additionally, a more complex model was designed, simulating a cartilage lesion (Fig. 4), and the reproducibility of the algorithm was assessed, as described previously.

In Vivo Reproducibility of 3D Cartilage Thickness Measurements

To determine the reproducibility of quantitative 3D cartilage thickness measurements in the living, we used MRI data sets of the human knee joint in which the repeatability of cartilage volume measurements and cartilage thickness maps had been analyzed previously (20,21). In these two studies, the knees of eight healthy volunteers were imaged with a 1.5-T magnet (Magnetom VISION; Siemens, Erlangen, Germany), using previously validated high-resolution, fat-suppressed 3D gradient-echo sequences (8–14,17–19). In one study (20), sagittal images of the entire knee joint had been obtained at a resolution of $2 \times 0.31 \times 0.31$ mm$^3$ (fat-suppressed fast low-angle shot (FLASH); TR 60 msec, TE 11 msec, flip angle 30°, number of excitations (NEX) 1, imaging time 20 min); in the other study (21), transverse images of the patella were obtained, with a sequence that had been optimized for a fast acquisition time of 4 hr 10 min (resolution $2 \times 0.58 \times 0.58$ mm$^3$ interpolated by zero filling before fast Fourier transforma-
tion to $2 \times 0.29 \times 0.29$ mm$^3$; TR 43 msec, TE 6 msec, flip angle 30°, NEX 1) (Fig. 2a). Six data sets were obtained of each volunteer, the knees being moved and repositioned in the coil between replicated examinations. The data were then transferred to an SGI workstation (Silicon Graphics, Mountain View, CA) and the 3D EDT algorithm described above was applied to each of the data sets. The interscan reproducibility of the mean and maximal patellar and tibial cartilage thickness was assessed by calculating the relative standard deviation [coefficient of variation (CV)%: standard deviation divided by the mean $\times 100$] of the sixfold determination of each cartilage plate in the eight volunteers (40 degrees of freedom). These values were compared with the reproducibility obtained for the cartilage volumes as described in the two previous studies (20,21). The average reproducibility was computed based on the root-mean-square average of the individual relative standard deviations (25), rather than the arithmetic mean of the eight values, as the latter may lead to slight overestimations of the true methodological precision (25). Additionally, we calculated the reproducibility of the standard deviation (in % of the mean thickness) of all thickness values within one surface, as a measure of the thickness inhomogeneity of each cartilage plate (15). The biological variation of the thickness data was estimated by calculating the mean value and the relative standard deviation across the eight volunteers, and these values were related to the average methodological variation, because the discriminative power of the method in cross-sectional studies depends on the ratio between these two values.

RESULTS

Test Models for the 3D Euclidian Distance Algorithm

When the analytical solution for the test models was compared with the 3D EDT algorithm, the maximal thickness values were identical, and the differences of the mean thickness values due to the discrete sampling of the test models were about 0.1 voxels. In the rotated and translated models, the maximal deviation of the mean thickness value from the analytical solution in any of the 10 positions was 0.35 voxels for the cylinder and 0.59 voxels for the ramp model. The deviations of the maximal thickness value were 0.11 and 0.26 voxels, respectively. These values correspond to 0.11 and 0.18 mm for the mean, and to 0.03 and 0.08 mm for the maximal thickness, if a suggested in-plane resolution for cartilage imaging of about 0.3 mm is assumed. The CV% of replicated thickness calculations at various orientations was 0.64% and 0.72% for the mean thickness of the cylinder and ramp model, and 0.12% and 0.27% for the maximal thickness. The reproducibility of the thickness computation in the model with the simulated cartilage lesion (Fig. 4) was 0.59% for the mean and 0.12% for the maximal thickness for repositioning. Measuring normal to the bone-cartilage interface instead of normal to the cartilage surface in the model with the simulated defect produced somewhat different results (0.37 voxels difference for the averaged mean thicknesses) (Fig. 4), but no significant differences in terms of reproducibility.
In Vivo Reproducibility of 3D Cartilage Thickness Measurements

Analysis of the six repeated examinations (with repositioning) in the eight healthy volunteers showed that the mean cartilage thickness computed with the 3D EDT algorithm yielded a similar degree of reproducibility as the cartilage volume determined in the two previous investigations (20,21: sagittal imaging of the knee at high resolution and transverse imaging of the patella at low acquisition time). The maximal thickness values and the relative standard deviation, describing the thickness inhomogeneity in the individual joint surfaces, generally yielded a somewhat lower degree of precision. As a quantitative measure of the discriminative power of the technique in cross-sectional studies, the ratio of the technical reproducibility of the method (Reproducibility in Tables 1, 2) to the biological variation (Variability in Tables 1, 2) showed that generally the biological variability was substantially higher than technical reproducibility. This ratio was about 15:1 for the mean patellar cartilage thickness and 5:1 for the medial tibia. For maximal thickness, it was 7:1 in the patella and about 2:1 in the medial tibia (Tables 1, 2).

DISCUSSION

In this work we have developed an algorithm for statistically accurate 3D determination of the mean and maximum thickness of the articular cartilage, based on 3D Euclidean distance transformation, and we have assessed the reproducibility of MRI-based cartilage thickness measurements in healthy volunteers. The algorithm has been shown to produce valid results in various geometric test objects, and a high reproducibility of MRI-based thickness measurements could be demonstrated.

Computational Method

Approaches to cartilage thickness measurements from MR image data have recently been investigated by several research groups. Robson et al (26) and Solloway et al (27) focused on the automatic delineation of the cartilage from the surrounding tissue to improve the segmentation process. To provide an optimal contrast on both the bone-cartilage and the cartilage-synovial interface, Robson et al (26) combined two imaging sequences for the automatic delineation of cartilage boundaries, using an edge detection algorithm. However, the distances between the cartilage contours were only assessed within the 2D imaging plane and were therefore dependent on the slice position and orientation. This may lead to important measuring errors due to different orientation of the joint relative to the image plane in cross-sectional and longitudinal studies, as these cannot be positioned identically. The approach of Solloway et al (27) suffers from the same limitation. A deformable cartilage model was adapted to the image data in a multi-resolution search, and the thickness was measured normal to the medial axes of both contours. Even if the algorithm was shown to perform well, no accurate quantification of the cartilage thickness in the MR images was possible due to the relative low in-plane resolution of the imaging sequence employed (0.89 x 0.89 mm$^2$) compared with the resolution used in this study (0.31 x 0.31 mm$^2$).

As in our approach, Ateshian et al (15) and Cohen et al (19) overcome the deficiencies of out-of-plane deviations of the measuring vectors by reconstructing the 3D articular topography of the cartilage. In contrast to our method, in which the cartilage thickness is determined as the minimal 3D Euclidean distance from data point to data point,

\[ f(x, y) = \sqrt{x^2 + y^2} - r \]

\[ f(x, y) = b \sin(\omega x) \sin(\omega y) + ax + c \]
without explicit calculation of normal vectors, Athesian et al (15) and Cohen et al (19) fitted a B-spline surface representation to the image data from which the thickness was calculated as the length of the normal vectors originating on the bone-cartilage interface and pointing to the cartilage surface. However, the high-resolution stereophotogrammetry technique employed by Athesian et al (19) (suggested accuracy 90 μm) is destructive and can only be applied in vitro. In the MRI study of Cohen et al (15), only the central parts of cadaveric knee joint cartilages were imaged, and the reproducibility of the method was not assessed in the living, in whom artifacts may occur due to blood flow or movement during image acquisition.

From a more theoretical point of view, our distance transformation method is highly efficient in terms of computational complexity for the calculation of distance maps in digital images (28). This aspect becomes more important when the method is introduced into clinical routine, where the time efficiency of the calculations is critical.

The evaluation of the method on geometric test models demonstrates that the deviations from the analytical results are not due to deficiencies of the algorithm, but are related to discretization errors during the sampling of the rotated and translated surfaces. However, the errors of about 0.1 mm are far below the image resolution and are small compared with the measuring errors that are expected to be introduced by partial volume effects and segmentation. The reproducibility of 0.12–0.72% for the mean thickness values at repositioning demonstrates a high precision and robustness of the algorithm.

Table 1
Analysis of the Knee Joint Cartilage Thickness With the 3D EDT Algorithm: Sagittal Knee Images*

<table>
<thead>
<tr>
<th>Cartilage volume</th>
<th>Mean thickness</th>
<th>Maximal thickness</th>
<th>Standard deviationb (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patella</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean</td>
<td>3.51 mL</td>
<td>2.8 mm</td>
<td>5.9 mm</td>
</tr>
<tr>
<td>Variability</td>
<td>20%</td>
<td>20%</td>
<td>19%</td>
</tr>
<tr>
<td>Reproducibilitya</td>
<td>1.5%</td>
<td>2.5%</td>
<td>2.9%</td>
</tr>
<tr>
<td>Medial tibia</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean</td>
<td>2.33 mL</td>
<td>1.6 mm</td>
<td>3.5 mm</td>
</tr>
<tr>
<td>Variability</td>
<td>20%</td>
<td>17%</td>
<td>12%</td>
</tr>
<tr>
<td>Reproducibilitya</td>
<td>3.2%</td>
<td>3.4%</td>
<td>7.9%</td>
</tr>
<tr>
<td>Lateral tibia</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean</td>
<td>2.79 mL</td>
<td>2.2 mm</td>
<td>4.5 mm</td>
</tr>
<tr>
<td>Variability</td>
<td>16%</td>
<td>16%</td>
<td>16%</td>
</tr>
<tr>
<td>Reproducibilitya</td>
<td>3.8%</td>
<td>2.8%</td>
<td>4.2%</td>
</tr>
</tbody>
</table>

*Comparison with the precision of cartilage volume measurements (20) in eight healthy volunteers based on a high-resolution fat-suppressed gradient-echo sequence.

The average reproducibility is measured as root mean square of the individual relative standard deviation (coefficient of variation) of the eight volunteers instead of the arithmetic mean (23).

$^a$Mean standard deviation of the thickness distribution in % of the mean value as a measure of the inhomogeneity of the cartilage plate.

In Vivo Reproducibility

Obviously, in a clinical context the precision of cartilage thickness measurements does not only depend on the accuracy of the computational algorithm, but also on the quality of the image data itself. For this reason, we applied the algorithm to replicated MR examinations of knee joints of healthy volunteers, to estimate the interscan reproducibility of the method. In comparison with the reproducibility reported for cartilage volume measurements by Peterfy

Table 2
Analysis of the Knee Joint Cartilage Thickness With the 3D EDT Algorithm: Transverse Patella Images*

<table>
<thead>
<tr>
<th>Cartilage volume</th>
<th>Mean thickness</th>
<th>Maximal thickness</th>
<th>Standard deviationb (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patella</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean</td>
<td>4.85 mL</td>
<td>2.9 mm</td>
<td>5.8 mm</td>
</tr>
<tr>
<td>Variability</td>
<td>29%</td>
<td>20%</td>
<td>20%</td>
</tr>
<tr>
<td>Reproducibilitya</td>
<td>1.6%</td>
<td>1.5%</td>
<td>2.1%</td>
</tr>
</tbody>
</table>

*Comparison with the precision of cartilage volume measurements (21) in eight healthy volunteers based on a fast fat-suppressed gradient echo sequence.

The average reproducibility is measured as root mean square of the individual relative standard deviation (coefficient of variation) of the eight volunteers instead of the arithmetic mean (23).

$^b$Mean standard deviation of the thickness distribution in % of the mean value as a measure of the inhomogeneity of the cartilage plate.
et al (8) and Marhsall et al (9), and with those reported in our two previous studies (20,21), the precision of the mean cartilage thickness was found to be in the same range, whereas that of the maximal thickness and the thickness variation within each surface was somewhat less. It is obvious that the maximal thickness is less reproducible than the mean cartilage thickness, because the maximal value depends on a single measurement (rather than averages of multiple measurements) and is therefore more highly affected by variations in MR image position and orientation as well as by the result of the local image segmentation. It should be noted that the reproducibilities given for the volume measurements differ somewhat from our previous papers (20,21), because in the current analysis the root-mean-square average (and not the arithmetic mean) of the individual relative standard deviations was computed. Using this definition, it has been shown (25) that for 40 degrees of freedom (=8 volunteers x 6 repetitions) the real reproducibility in the population should not be overestimated by more than 25% at a confidence level of 95%.

To our knowledge the current work is the first to report a high reproducibility of 3D quantification of the knee joint cartilage thickness in the living with MRI, independent of the original section position and orientation. It should be kept in mind that these measurements cannot be made from serial MR images alone (without 3D reconstruction and digital postprocessing), as the angulation of the image plane with the articular surface varies throughout the joint and cannot be known a priori.

Reproducibility in the patella was found to be higher than that in the tibia, and the same holds for the discriminative power of the method in cross-sectional studies, which can be estimated based on the ratio of the technical reproducibility to the biological variation (the values of Reproducibility vs. Variability in Tables 1 and 2). Generally, we found the biological variation to be substantially higher than the technical precision. Whereas in the patella the ratios of the technical reproducibility to the biological variation of about 15:1 were obtained for the mean, and 10:1 for the maximal thickness, these ratios did attain smaller values in the tibia. This is most probably due to the lower absolute cartilage thickness in the tibia compared with the patella, and to higher partial volume effects in sagittal images. Using a higher in-plane resolution and/or another section orientation in the MR images may help to increase these ratios, by improving the accuracy of the segmentation process. Overall, the discriminative power of the volume measurements was slightly higher than that of the thickness measurements, this being caused by an presumably higher biological variability of the cartilage volume and a slightly better reproducibility.

In this context it should be mentioned that the data reported here are valid for healthy cartilage. Further studies will be required to determine the reproducibility of the method at various stages of osteoarthritic change, as under these conditions partial volume effects may become higher. Particularly in joint degeneration, however, the analysis of the cartilage thickness may provide more comprehensive data than that of the volume, as the possibility also exists to monitor focal cartilage lesions, which may affect the thickness at a specific location, but not so much the overall amount of cartilage tissue present. Moreover, patient-specific computer models of diarthrodial joints (e.g., 2, 3), in which the effect of an operation can be planned and optimized, require precise information on cartilage thickness (and not volume) for the accurate computation of joint contact areas and contact stresses.

The precision of the technique presented may be further increased by improving the accuracy of the interactive segmentation process. Therefore, in future work, we plan to devote efforts to the automatization of the delineation of the cartilage. To avoid undersampling in the z-direction and to provide more isotropic voxels, the ratio of the slice thickness to in-plane resolution should be reduced by decreasing the slice thickness. Moreover, the technique should be extended to the quantitative comparison of 3D cartilage thickness maps, by matching the cartilage surfaces and subtracting corresponding cartilage thickness values from subsequent examinations of the same individual.

ACKNOWLEDGMENTS

Results of this study were obtained as part of the doctoral thesis of Tobias Stammberger at the Ludwig-Maximilians-Universität München (in preparation).

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