A Novel Method for Boundary Detection and Thickness Measurement of Two Adjacent Thin Structures from 3-D MR Images

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SUMMARY To determine the thickness from MR images, segmentation, that is, boundary detection, of the two adjacent thin structures (e.g., femoral cartilage and acetabular cartilage in the hip joint) is needed before thickness determination. Traditional techniques such as zero-crossings of the second derivatives are not suitable for the detection of these boundaries. A theoretical simulation analysis reveals that the zero-crossing method yields considerable biases in boundary detection and thickness measurement of the two adjacent thin structures from MR images. This paper studies the accurate detection of hip cartilage boundaries in the image plane, and a new method based on a model of the MR imaging process is proposed for this application. Based on the newly developed model, a hip cartilage boundary detection algorithm is developed. The in-plane thickness is computed based on the boundaries detected using the proposed algorithm. In order to correct the image plane thickness for overestimation due to oblique slicing, a three-dimensional (3-D) thickness computation approach is introduced. Experimental results show that the thickness measurement obtained by the new thickness computation approach is more accurate than that obtained by the existing thickness computation approaches.

key words: boundary detection, cartilage, magnetic resonance imaging, osteoarthritis, segmentation

1. Introduction

Magnetic resonance (MR) imaging allows direct visualization of the articular cartilage in the 3D spaces, due to its excellent soft tissue contrast and ability to acquire 3-D information. The 3D distribution of the articular cartilage in other joints, such as knee, shoulder [1], elbow [2], ankle and hind foot [3], has been studied quantitatively using MR imaging in conjunction with computational processing techniques. In recent investigations, the articular cartilage of the hip joint has been visualized using MR imaging [4].

A cartilage thickness measurement technique using MR imaging is increasingly being used for the diagnostics of pathologies related to alterations in this tissue. The lack of studies concerning cartilage thickness measurement is mainly due to the difficulty of imaging the articular cartilage in the hip joint. Because the spacing between acetabular and femoral cartilage is very narrow, it is difficult to discriminate between these two tissues. In order to discriminate the acetabular cartilage from the femoral head cartilage, a traction device was used on the legs during image acquisition [5]. Several investigators have developed thickness measurement techniques that generate a cartilage thickness map along the radial directions originating from the center of the femoral head [5]–[8]. In these studies, the inner and outer cartilage edge positions were detected using the first and second directional derivatives along radial directions originating from the center of the femoral head. The thickness of the femoral and acetabular cartilage was defined as the distance between the inner and outer cartilage edge positions along the radial directions.

In our previous study [9], results show that the conventional edge methods (e.g., the first and second directional derivatives) can introduce large measurement errors in thickness measurement of the hip joint cartilage due to the intrinsic limitations in the finite spatial resolution. In another study, Wyle et al. [10] performed the cartilage thickness measurement in MR-arthrography cadaver hips, with coronal anatomic slice as gold standard. They calculated the mean cartilage-thickness values on anatomic coronal slice and MR-arthrographic section through the center of the femoral head. They found that the thickness of the femoral and acetabular cartilage was underestimated compared to anatomic thickness in the coronal plane. For instance, acetabular cartilage thickness values were estimated by 2.97 ± 0.28 mm for anatomic measurement and 2.56 ± 0.67 mm for MR measurement on the superolateral part; and femoral cartilage thickness was estimated by 1.85 ± 0.15 mm for anatomic measurement and 1.38 ± 0.38 mm for MR measurement on the superomedial part.

The present study aims at the development of a novel framework, named the model-based method, designed for improving the accuracy in boundary detection of the human hip cartilage. Further, a set of image processing modules to compute hip cartilage thickness was also developed. The edge detection approach presented in this paper is specifically developed for MR images of the hip joint cartilages but is applicable to other complicated thin structures (more than one material) sites as well. For example, pieces of the brain cortex are curved and nearby each other. They have similar gray values in MR images. Intracranial blood vessels are nearby the bones. Both bones and blood vessel appear with
similar density in CT angiography images. In other words, they both have similar gray values. In these cases, the adjacent structures influence each other. Traditional segmentation methods often fail in segmenting the two adjacent thin structures with similar gray values. The new segmentation method in this paper can be used for the above situations.

Using 5 normal cadaver hips, we present results showing that the proposed method is more accurate than the conventional edge detection method for boundary detection and thickness measurement of hip cartilages. Furthermore, 15 hips of patients with OA are also used to demonstrate the clinical applicability of our proposed method.

The contribution of this work consists of the following.

1) We formulate a theoretical simulation method to evaluate the boundary detection accuracy of two adjacent thin structures in 3-D MR images. Two adjacent thin structures, MR imaging processes, and post-processing for boundary detection are modeled and simulated. It provides an easy way to test the segmentation accuracy of existing methods from 3-D MR images.

2) We present a new method based on a model of the MRI process for accurately detecting the boundaries of two adjacent thin structures and determining their thickness in a 2-D image plane. It opens the road for overcoming the inherent limits on segmentation accuracy arising from finite resolution.

3) In order to correct the in-plane thickness for overestimation due to oblique slicing, a 3-D thickness computation approach is introduced. It applies a fast, simple, and easily conceivable algorithm to solve the accurate thickness measurement task in 3-D MR images. This can be extended to the accurate boundary detection in 3-D MR images.

2. Theoretical Simulation

2.1 Modeling Two Cartilages of Hip Joint

Because the femoral head typically has a spherical shape, a spherical coordinate system is defined for the hip joint (Fig. 1(a)). The center of the femoral head corresponds to the center of the spherical coordinate system. For simplification of the coordinate system, images of right hips are converted to their mirror images (i.e., left hip) by mirroring with respect to the mid-sagittal plane of the body. In this spherical coordinate system, the location of the acetabular (Fig. 1(b)) and femoral cartilage (Fig. 1(c)) is represented by spherical coordinate system, the location of the acetabular converted to their mirror images (i.e., left hip) by mirroring.

2.2 Modeling MR Image Acquisition

The one dimensional (1-D) point spread function (PSF) of MR images is given by [11]

\[
m(x; \Delta_x) = \frac{1}{N_x} \sin(\pi \frac{x}{\Delta_x})
\]

(5)

where \(N_x\) is the number of samples in the frequency domain, and \(\Delta_x\) represents the sampling interval in the spatial domain, Eq. (5) is well-approximated by

\[
m(x; \Delta_x) = S inc(x; \frac{1}{\Delta_x})
\]

(6)

The 3-D PSF is given by

\[
m(\bar{x}; \Delta_x, \Delta_y, \Delta_z) = m(x; \Delta_x) \cdot m(y; \Delta_y) \cdot m(z; \Delta_z)
\]

(7)

where \(\Delta_x, \Delta_y, \Delta_z\) are sampling intervals along the x axis,
Fig. 1 (a) A celestial spherical coordinate system for a left hip. The enlarged representation of (b) the acetabulum and (c) the femoral head. Articular cartilage position is defined by longitude $\phi$ and latitude $\theta$ angles in the spherical coordinate.

Fig. 2 (a) Diagram showing the coronal plane through the femoral head center. C stands for the center of the femoral head. $L_s$, $L_{sb}$, and $L_0$ denote the ideal intensity distributions of the two cartilages (acetabular cartilage and femoral cartilage), the background of both sides and the joint space between the two cartilages, respectively. $L_{b-}$ is the ideal intensity distribution of the left side of a single thin structure (femoral cartilage without adjacent acetabular cartilage). Assuming that $STR_{fc}$ and $STR_{ac}$ represent the femoral cartilage and acetabular cartilage in the hip joint, respectively. (b) 1-D profile of the ideal intensity distribution of a single thin structure along the normal orientation of the thin structure. $\tau_{fc}$, $\tau_{ac}$, and $w$ are $STR_{fc}$, $STR_{ac}$ thickness, and distance between them, respectively. (c) 1-D profile of the ideal intensity distribution of the two adjacent thin structures along the normal orientation of the thin structure. $\tau$ is the thickness of the thin structure.

In actual MRI, the magnitude operation is applied to the complex number obtained at each voxel by FFT reconstruction, whose effects are not negligible. Thus, the MR image of two adjacent thin structures with orientation $\hat{r}_{\theta,\phi}$ is given by

$$f(\vec{x}; x_1, x_2, x_3, x_4) = |S(\vec{x}; x_1, x_2, x_3, x_4, \hat{r}_{\theta,\phi})| * m(\vec{x}; \Delta x, \Delta y, \Delta z)$$  \hspace{1cm} (8)

where * denotes the convolution operation.

Effects of MR image and limited in-plane resolution can be elucidated by analyzing 1-D profile of MR intensity values, $f(\vec{x}; x_1, x_2, x_3, x_4)$, along the straight line (the normal orientation) given by

$$\vec{x} = s \cdot \hat{r}_{\theta,\phi}$$ \hspace{1cm} (9)

where $s$ is a parameter representing the position on the straight line. By substituting Eq. (9) for $\vec{x}$ in $f(\vec{x}; x_1, x_2, x_3, x_4)$,

$$f(s; x_1, x_2, x_3, x_4) = f(s \cdot \hat{r}_{\theta,\phi}; x_1, x_2, x_3, x_4, \hat{r}_{\theta,\phi})$$ \hspace{1cm} (10)
Figure 3 shows the 1-D profiles of ideal intensity and simulated MR signal intensity along a line perpendicular to the thin structure for two adjacent thin structures (a) and a single thin structure (b).

2.3 Boundary Detection and Thickness Measurement Based on the Zero-Crossing Method

Zero-crossings of the second derivatives along the gradient direction were introduced by Haralick [12] and then used by Canny [13] as the edge detectors. Since its publication, zero-crossing method has become one of the most active and successful research areas in edge detection. In this section, the zero-crossing method was used for illustrating potential error in the edge detection and thickness measurement of two adjacent thin structures. The image edges are defined as the zero-crossings of the second directional derivatives along the normal direction of thin structure, which is equivalent to the Canny edge detector [13]. Gaussian blurring is typically combined with the second directional derivatives to adjust scale as well as reduce noise.

The partial second derivative combined with Gaussian blurring for the MR image \( f(\vec{x},x_1,x_2,x_3,x_4) \), for example, is given by

\[
f_{xx}(\vec{x},x_1,x_2,x_3,x_4,\sigma) = f(\vec{x},x_1,x_2,x_3,x_4)
\]

\[
\ast g_{xx}(\vec{x};\sigma)
\]

where

\[
g_{xx}(\vec{x};\sigma) = \frac{\partial^2}{\partial x^2} \text{Gauss}(\vec{x};\sigma)
\]

in which Gauss(\( \vec{x};\sigma \)) is the isotropic 3D Gaussian function with Standard Deviation (SD) \( \sigma \) given by

\[
\text{Gauss}(\vec{x};\sigma) = \frac{1}{\sqrt{(2\pi)^3\sigma^3}} e^{-(x^2+y^2+z^2)/2\sigma^2}
\]

The second directional derivative along \( \vec{r}_{\theta,\phi} \) is represented as

\[
f''(\vec{t},x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi}) = g_{xx}(\vec{t};\sigma)
\]

\[
\ast f(\vec{t},x_1,x_2,x_3,x_4)
\]

where \( \vec{t} = R_{\theta,\phi}\vec{x} \), in which \( R_{\theta,\phi} \) denotes a 3 \( \times \) 3 matrix representing rotation which causes the normal orientation of the thin structure \( S_0(\vec{x},x_2,x_3,x_4) \), i.e., the \( x \) axis, to correspond to \( \vec{r}_{\theta,\phi} \). Similarly, the first directional derivative along \( \vec{r}_{\theta,\phi} \) is represented as

\[
f'(\vec{t},x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi}) = g_{xx}(\vec{t};\sigma)
\]

\[
\ast f(\vec{t},x_1,x_2,x_3,x_4)
\]

The boundary position of thin structure can be localized by analyzing 1-D profile of \( f''(\vec{t},x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi}) \) and \( f'(\vec{t},x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi}) \) along the normal orientation \( \vec{r}_{\theta,\phi} \). By substituting Eq. (9) for \( \vec{t} \) in \( f''(\vec{t},x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi}) \) and \( f'(\vec{t},x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi}) \),

\[
f''(s) = f''(s \cdot \vec{r}_{\theta,\phi};x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi})
\]

and

\[
f'(s) = f'(s \cdot \vec{r}_{\theta,\phi};x_1,x_2,x_3,x_4,\sigma,\vec{r}_{\theta,\phi})
\]

are derived, respectively.

To illustrate potential error in the edge detection of two adjacent thin structures when applying the zero-crossing method, we give some examples for detecting the two edges of the \( ST_{ac} \). Figure 4(a) shows the edge detection procedure of the \( ST_{ac} \) using zero-crossings of the second directional derivatives along the normal direction. We obtained zero-crossing points \( s = q \) and \( s = p \) on the left and right sides of the edges for the \( ST_{ac} \), which satisfy \( f''(s) = 0 \).

Let zero-crossing points \( q \) and \( p \) correspond to the minimum and maximum values of \( f'(s) \) among those satisfying the condition given by \( f''(s) = 0 \). The two sides of the \( ST_{ac} \) edge are localized at zero-crossing positions \( q \) and \( p \). The measured thickness of the \( ST_{ac} \), \( T_{ac} \), is defined as the distance between the two detected boundary points, which is given by

\[
T_{ac} = |p - q|
\]

In Fig. 4(a), when \( w = 0.5 \) mm, the true thickness of 3.0 mm is measured to be approximately 2.63 mm (a 12.33% underestimate of the true thickness of 3.0 mm) in the 2-D slice MR image. Figure 4(b) shows the measured thickness of a single thin structure. In Fig. 4(b), the true thickness of 3.0 mm is measured to be approximately 3.03 mm (a 1.0% overestimate of the true thickness of 3.0 mm) in the 2-D image plane. Therefore, the thickness of a single thin structure is accurately measured, while the thickness of the \( ST_{ac} \) is underestimated.

Figure 5 shows the boundary positions and the measured thickness values when \( w = 1.5 \) mm in the image plane. The true thickness of 1.5 mm is measured to be approximately 1.55 mm (3.3% error) in the image plane. So, the measured thickness values are a good approximation of their true values under those measured values satisfying the condition of \( T_{fc} \geq 1.5 \) mm, \( W \geq 1.5 \) mm and \( T_{ac} \geq 1.5 \) mm.
Thickness of the thin structure measured by the zero-crossing method in the $x$-$y$ plane. The in-plane resolution: $\Delta_x = \Delta_y = \Delta_{xy} = 0.625$; Gaussian standard deviation: $\sigma = (2^{1/2}/2)\Delta_{xy}$; $\tau_{fc} = 2$ mm, $w = 0.5$ mm, $\tau_{ac} = 3.0$ mm. (a) Two adjacent thin structures. Measured thickness of the STR$_{ac}$: $T_{ac} = |p - q|$. In the case of $w = 0.5$ mm, thickness of $\tau_{ac} = 3.0$ mm was measured by approximately 2.63 mm (-12.33% error). The edge detection method exhibited considerable underestimation error for the two adjacent thin structures. (b) A single thin structure. Measured thickness: $T = |p - q|$. Thickness of $\tau = 3.0$ mm was measured by approximately 3.03 mm (1.0% error).

Figure 4

Fig. 5  

The boundary positions and the measured thickness values in the case of the joint space width of $w = 1.5$ mm. The in-plane resolution: $\Delta_x = \Delta_y = \Delta_{xy} = 0.625$; Gaussian standard deviation: $\sigma = (2^{1/2}/2)\Delta_{xy}$; $\tau_{fc} = 1.5$ mm, $w = 1.5$ mm, $\tau_{ac} = 1.5$ mm. $T_{fc} (T_{fc} = |m - n|)$, $W (W = |q - m|)$ and $T_{ac} (T_{ac} = |p - q|)$ are the measured values of $\tau_{fc}$, $w$ and $\tau_{ac}$. In this figure, the boundary positions of $n$, $m$, $q$, and $p$ estimated by the zero-crossing method are approximately equal to the true boundary positions of $x_1$, $x_2$, $x_3$, and $x_4$.

For this case, the zero-crossing method can accurately localize the boundaries of the STR$_{fc}$ and the STR$_{ac}$, i.e., the boundary positions of $n$, $m$, $q$, $p$ defined by the zero-crossing method are approximately equal to the true boundary positions of $x_1$, $x_2$, $x_3$, and $x_4$.

Figure 6 shows the measured thickness values in the 3-D MR images. In Fig. 6, thickness of $\tau_{ac} = 3$ mm was measured by approximately 2.56 mm (a 14.67% underestimate of the true thickness of 3.0 mm) for $\varphi = 0^\circ$ and $\theta = 45^\circ$ (Fig. 6(a)), and 2.15 mm (a 28.33% underestimate of the true thickness of 3.0 mm) for $\varphi = 0^\circ$ and $\theta = 90^\circ$ (Fig. 6(b)). It should be noted that the thickness measurement values varied with variable normal orientation of the thin structure.

3. Materials and Methods

Four acrylic plate phantoms of sheet-like objects with known thickness, five fresh-frozen cadaver hips and fifteen hips of patients with OA were examined. The method proposed in this paper is performed in two steps. First, edge detection and thickness measurement in the scan plane are performed based on a model of the MRI process. Subsequently, cartilage thickness overestimation is corrected due to oblique slicing in the 3-D images.

3.1 Sample Preparation and Image Processing

3.1.1 Acrylic Plate Phantoms

We imaged the four acrylic plate phantoms of sheet-like objects with known thickness. One was used for a single thin structure; the other three were used for two adjacent thin structures. The single thin structure phantom was used to compare with the thin structure influenced by the adjacent thin structure.

The phantom for the single thin structure consisted of four acrylic plates of 80 x 80 mm$^2$ with thickness of $\tau = 2.0$, 1.0, 1.5, and 3.0 mm, placed parallel to each other.
Fig. 6  Thickness measurement procedure using zero-crossings of the second directional derivatives along the normal direction. Graph shows the thickness measurement of the STR_{ac}. The voxel size: Δ_{xy} = 0.625 (mm), Δ_{z} = 1.5 (mm). Gaussian standard deviation: σ = (2^{1/2}/2)Δ_{xy}. MR intensity: L_s = 100, L_b+ = 10, L_0 = 15. Joint space width: w = 0.5 mm; thickness of the thin structure: τ_{fc} = 2 mm, and τ_{ac} = 3 mm. Measured thickness of the STR_{ac}: T_{ac} = |p − q|. (a) ϕ = 0° and θ = 45°. Thickness of τ_{ac} = 3 mm was measured by approximately 2.56 mm (-14.67% error). (b) ϕ = 0° and θ = 90°. Thickness of τ_{ac} = 3 mm was measured by approximately 2.15 mm (-28.33% error).

Fig. 7  Acrylic plate phantoms. (a) Physical appearance of a single thin structure. (b) MR image of the 3-D two adjacent acrylic plates (STR_{fc} and STR_{ac}). It is composed of four pairs of acrylic plates, placed parallel with each other. The interval between the STR_{fc} and the STR_{ac}: w = 0.5 mm. Graph shows the thickness of the acrylic plates. (c) MR image of the 2-D two adjacent acrylic plates (STR_{fc} and STR_{ac}). The interval between the STR_{fc} and the STR_{ac}: w = 1.5 mm. Graph shows the thickness of the acrylic plates.

(Fig. 7(a)). Each of the three phantoms for two adjacent thin structures consisted of four pairs of acrylic plates, placed parallel to each other. One of their volume rendering images is shown in Fig. 7(b). The parameters (τ_{fc}, w and τ_{ac}) of three phantoms are given in Table 1(a) for w = 0.5 mm, Table 1(b) for w = 1.0 mm, and Table 1(c) for w = 1.5 mm (Fig. 7(c)).

All phantoms were imaged with different normal positions of the phantom plates, eight with variable θ (θ = 0°, 15°, 25°, 35°, 45°, 60°, 75°, and 90°) and fixed ϕ (ϕ = 0°), and five with variable ϕ (ϕ = 0°, 15°, 25°, 35°, and 45°) and fixed θ (θ = 0°). We compared actually measured thickness from the real MR data with the computational thickness calculated by the numerical simulations.

3.1.2 Cadaver Hip Specimens for Anatomic Measurement

3 fresh-frozen cadaver specimens of normal human hip joints were examined. To obtain a “gold standard” measurement of the JSW and cartilage thickness, we designed
Table 1 Phantom parameters for two adjacent thin structures. Table shows $\text{STR}_{fc}$ thickness ($\tau_{fc}$), $\text{STR}_{ac}$ thickness ($\tau_{ac}$), and distance between $\text{STR}_{fc}$ and $\text{STR}_{ac}$ ($w$).

(a) Distance between $\text{STR}_{fc}$ and $\text{STR}_{ac}$: $w = 0.5$ mm.

| Thin structure number | $\tau_{fc}$ | $w$ | $\tau_{ac}$ |
|-----------------------|-------------|-----|-------------|
| 1                     | 1.5 mm      | 0.5 mm | 1.0 mm      |
| 2                     | 1.5 mm      | 0.5 mm | 1.5 mm      |
| 3                     | 1.5 mm      | 0.5 mm | 2.0 mm      |
| 4                     | 1.5 mm      | 0.5 mm | 3.0 mm      |

(b) Distance between $\text{STR}_{fc}$ and $\text{STR}_{ac}$: $w = 1.0$ mm.

| Thin structure number | $\tau_{fc}$ | $w$ | $\tau_{ac}$ |
|-----------------------|-------------|-----|-------------|
| 1                     | 1.5 mm      | 1.0 mm | 1.0 mm      |
| 2                     | 1.5 mm      | 1.0 mm | 1.5 mm      |
| 3                     | 1.5 mm      | 1.0 mm | 2.0 mm      |
| 4                     | 1.5 mm      | 1.0 mm | 3.0 mm      |

(c) Distance between $\text{STR}_{fc}$ and $\text{STR}_{ac}$: $w = 1.5$ mm.

| Thin structure number | $\tau_{fc}$ | $w$ | $\tau_{ac}$ |
|-----------------------|-------------|-----|-------------|
| 1                     | 1.5 mm      | 1.5 mm | 1.0 mm      |
| 2                     | 1.5 mm      | 1.5 mm | 1.5 mm      |
| 3                     | 1.5 mm      | 1.5 mm | 2.0 mm      |
| 4                     | 1.5 mm      | 1.5 mm | 3.0 mm      |

an experimental procedure similar to the technique used by McGibbon et al. [14], and then performed the comparisons among MR measurements obtained with the zero-crossing method, MR measurements obtained with the proposed method and anatomic measurements.

Experimental setup consisted of the following five steps (Fig. 8(a)): (a) cutting the femoral neck perpendicular to the femoral neck; (b) marking the center of the femoral neck; (c) locating the femoral head center and the anatomic coronal slice through the femoral head center (literature [10] describes a detailed procedure for locating the femoral head center and anatomic coronal slice); (d) making the bony defects artificially in the pelvis and femur for landmarks of the anatomic coronal slice; and (e) drilling a hole from the femoral neck center to the femoral head center.

The specimen was mounted in a custom-made container that could be oriented at any desired angle about a femoral neck midline (drill line). Cross hairs engraved on the container lid allowed precise alignment of the specimen with the MR imaging plane. Assuming that one of the cross hairs precisely corresponded with the position of the coronal slice landmarks (Fig. 8(b)).

For each of the three specimens, anatomic measurements were performed in the coronal plane only (Fig. 9(a)); and two imaging series were acquired at two successively spaced orientations of the specimen: (a) through the anatomic coronal slice (Fig. 9(b)), and (b) perpendicular to the anatomic coronal slice (Fig. 9(c)). Each series consisted of 38 slices. After performing the MR imaging, the hip joints were sectioned into halves along the landmarks of the anatomic coronal slice for performing the anatomical measurement (See [10] for the detailed procedure of anatomical measurement).

2 cadaver hip specimens were used for the study. In [15], Cohen et al. obtained very good accuracy (mean error, 0.31 mm) for cartilage thickness measurements of cadaveric knees using a B-spine curve fitting technique, and using stereophotogrammetric (SPG) measurements as the gold standard. Like prior study (Cohen’s study), we evaluated the accuracy of quantification in cartilage thickness from MRI relative to a “gold standard” SPG measurement. A detailed description for experimental setup of SPG measurement has been published previously [15].

In Section-IV, we perform the comparisons between MR measurements obtained with the proposed method and SPG measurements, and between MR measurements obtained with the zero-crossing method and SPG measurements.

3.1.3 Cadaver Hip Specimens for Stereophotogrammetric Measurement

15 hips of patients with OA (9 women, 6 men; age range, 37–65 years; mean age, 52.6 years) were used. In the living hips, both the femoral head and the acetabulum are covered with cartilage. The ball and socket constitution of the hip joint, with strong capsule and ligaments, does not permit discrimination of the articular cartilage of the femoral head from the acetabulum. The joint space width (JSW) between the femoral cartilage and acetabular cartilage is much narrower in the hip than in the knee, making it difficult to separate the cartilage on opposing articulating bone surfaces.

To allow clear separation of acetabular and femoral cartilage on MR images, the original continuous leg traction technique was used during MR imaging in the experiment [5]. Briefly, this system comprises a leg apparatus that pulls the leg caudally with approximately 15 kg of force,
Fig. 9  Diagrams showing the measurement points used in the center plane (marked coronal plane). (a) Anatomic coronal slice through the femoral neck midline (drill line). Make the bony defects artificially in the pelvis and femur for landmarks of the anatomic coronal slice. After performing the MR imaging, the hip joint was sectioned into halves from the position of the landmarks. The anatomic thickness of articular cartilage, measured by an expert at 10° increments, was compared with cartilage thickness on the corresponding MR images. (b) MR imaging plane through the center plane from the position of the landmarks. The included angle between the MR imaging plane and the center plane is 0°. The cartilage thickness was measured along the radial direction from the center of the femoral head at 10° increments (range, 10° ∼ 90°) of longitude $\phi$ with the fixed $\theta$ ($\theta = 0°$). (c) MR imaging plane perpendicular to the center plane. The specimen mounted in a custom-made container was rotated by 90° around the femoral neck midline. The included angle between the MR imaging plane and the center plane is 90°. The cartilage thickness was measured along the radial direction from the center of the femoral head at 10° increments (range, 10° ∼ 90°) of latitude $\theta$ with the fixed $\phi$ ($\phi = 0°$).

and a pelvic apparatus that pulls the pelvis cranially with approximately 10 kg of force; thus, traction force is exerted on the hip joint. Immediately after setting this traction device on patients, MR imaging was performed. Using this imaging technique, acetabular and femoral cartilages at high signal were clearly detected by interposition of low signal space between the two cartilages (Fig. 10). This space was enhanced in high signal by intravenous injection of gadolinium-DTPA in the study by Nishii et al. [5], and considered to represent joint fluid. Without use of the traction device during MR imaging, complete delineation of the cartilage border was difficult, due to the inherent adhesive nature of the two types of cartilage at the weight-bearing area.

3.1.5 MR Imaging and Image Processing

All MR imaging was performed with fat-suppressed 3-D fast spoiled gradient-echo (SPGR) sequence on a 1.5-T MR system (Horizon, General Electric). Imaging parameters were as follows: repetition time (TR)/echo time (TE), 24.4/5.7 ms; flip angle, 20°; section thickness = 1.5 mm; in-plane resolution, 0.625 mm; imaging matrix, 256 × 256. MR imaging with anisotropic (noncubic) voxels: $\Delta_{xy} = 0.625$ (mm), $\Delta_z = 1.5$ (mm). Thus, voxel anisotropy $=\Delta_z/\Delta_{xy} = 2.4$.

In the present study, the coordinate system is designated as follows: $x$ and $y$ axes were horizontal and vertical, respectively, in the image plane, and the $z$ axis was normal to the image plane. MR data are discrete samples from continuous images. Thus, interpolation of MR images is needed before thickness estimation. Our approach is to apply interpolation along the $x$-direction and $y$-direction. In this experiment, sinc interpolation was used along the $x$-direction and $y$-direction to make image the matrix size double in the frequency domain. The sampling interval in the interpolated data is 0.3125 ($=\Delta = (1/2)\Delta_{xy}$) mm in the $x$-, and $y$- directions.
3.2 Boundary Detection and Thickness Measurement Based on MR Imaging Model in 2-D

As described in the theoretical analysis, the zero-crossing method shows considerable bias in boundary location estimates of the two adjacent thin structures (Fig. 4(a)). To correct bias, we present a new method based on a model of the MR signal intensity. This is referred to as the “model-based method”. Similar to Eq. (10), the 1-D profile of the modeled MR signal intensity along the radial line \( s \) (the normal orientation \( \vec{r}_p \)) in the \( x-y \) plane, is given by

\[
f(s; x_1, x_2, x_3, x_4) = f(s; \vec{r}_p; x_1, x_2, x_3, x_4, \vec{r}_p) \quad (19)
\]

In our previous study [16], a detailed model-based procedure for accurate thickness measurement of two adjacent thin structures in 2-D image plane has been described. Similar to the procedure presented in [16], we can predict the shape of the MR signal intensity profile along the normal direction given in Eq. (19). The difference between the predicted and actual profiles observed in MR data is minimized by refining model parameters. The set of model parameters that minimizes the difference between model and actual MR data yield the boundary positions of \( x_1, x_2, x_3, \) and \( x_4 \). The femoral cartilage (\( STR_{fc} \)) thickness, the JSW between the femoral cartilage and acetabular cartilage and the acetabular cartilage (\( STR_{ac} \)) thickness are defined as \( r_{0fc} = x_2 - x_1 \), \( w_0 = x_3 - x_2 \) and \( r_{0ac} = x_4 - x_3 \), respectively.

Figure 11 indicates the determination procedures of the model-based method for estimating the boundary locations of the articular cartilage. In Fig. 11, to match the actual MR intensity profile with the predicted intensity profile model, continuous curved line model was discretized. In the estimation of cartilage boundary positions the model intensity profile \( f(s) \) as given in Eq. (19) is fitted to a measured intensity value profile along the radial line direction in the MR image shown in Fig. 10. The fit procedure yields estimates of positions \( x_1, x_2, x_3, \) and \( x_4 \) for the different cartilage layers (see Fig. 11).

3.3 3-D Thickness Measurement

The accurate estimate of cartilage thickness is obtained from the slice which intersects the joint normal to the cartilage surface (i.e., the center slice). All other slices intersect the surface obliquely and, when measured in the image plane, will overestimate cartilage thickness. 3-D thickness distribution of the acetabular cartilage is generated using the following method, which correct cartilage plane thickness values for overestimation due to oblique slicing.

In the \( j \)th slice, where slice index \( j = 0, 1, \ldots, n_s - 1 \) and \( n_s \) denotes the total number of slices, over small displacements both the cartilage surface and calcified interface of the acetabular are assumed concentric spheres (Fig. 12). The image plane thickness (\( r_{0ac} \)) at the point \( \vec{x}_i \) in the \( j \)th slice is determined by using \( r_{0ac}' = |x_i - x'_i| = |x_i - z_j| \), where \( \vec{x}_i = (x_{i1}, x_{i2}, x_{i3}, x_{i4}) = (x_i \cos \varphi, x_i \sin \varphi, n_j t_s) \), \( i=3, 4, n_s \) denotes the slice index, and \( t_s = (1.5 \text{ mm}) \) denotes the slice thickness.

The correction routine first locates the orientation angle \( (\alpha) \) of an auxiliary cutting plane which intersects the spherical joint normal to the cartilage surface at the point \( \vec{x}_i \). As shown in Fig. 12, the auxiliary plane is developed from the axial plane \( (x, z) \), and the auxiliary plane angle \( (\alpha) \) is referenced at the midpoint \( (\vec{q} = (q_x, q_y, q_z)) \) of the thickness vector \( (r_{0ac}) \).

\[
\alpha = \arctan((q_z - C_z)/(q_x - C_x)) \quad (20)
\]

where \( q_x = (x_{i1} + x_{i4})/2 = x_{i1} + x_{i2} \), \( C = (C_x, C_y, C_z) \) is the manually specified center position of the sphere approximating the femoral head.

In the axial plane projection, the radii \( r_{ac} \) of the endpoint positions of the thickness vector \( (r_{0ac}) \) are then calculated using
**Fig. 12** Graph illustrating a procedure for correcting the image plane thickness for overestimation due to oblique slicing in the 3-D images.

$$r_{x3} = \sqrt{(x_{3x} - C_x)^2 + (x_{3z} - C_z)^2}$$

$$= \sqrt{(x_3 \cos \varphi - C_x)^2 + (n_j t_x - C_z)^2}$$

$$r_{x4} = \sqrt{(x_{4x} - C_x)^2 + (x_{4z} - C_z)^2}$$

$$= \sqrt{(x_4 \cos \varphi - C_x)^2 + (n_j t_x - C_z)^2}$$

Assuming $\vec{x}_3$ and $\vec{x}_4$ are on concentric circles for small displacements allows calculation of the corrected endpoint coordinates $\vec{x}'_3$ and $\vec{x}'_4$.

$$x'_{3x} = C_x + r_{x3} \cos \alpha, x'_{3y} = x_{3y}, x'_{3z} = C_z + r_{x3} \sin \alpha$$

$$x'_{4x} = C_x + r_{x4} \cos \alpha, x'_{4y} = x_{4y}, x'_{4z} = C_z + r_{x4} \sin \alpha$$

Then the corrected thickness are calculated from

$$\tau'_{ac} = |\vec{x}'_4 - \vec{x}'_3|$$

$$= \sqrt{(x'_{4x} - x'_{3x})^2 + (x'_{4y} - x'_{3y})^2 + (x'_{4z} - x'_{3z})^2}$$

The process is then repeated for the image plane, which corrects for the surface obliquity slicing and gives true three-dimensional thickness measures. Similarly, the femoral cartilage thickness, $\tau'_{fc}$, can be calculated by the same determination procedure.

**4. Results**

4.1 Validating the Theoretical Simulation

Theoretical simulation based on the zero crossing method was performed for examining the effects of cartilage surface orientation ($\varphi$ and $\theta$), joint space width ($w$), and MR imaging resolution ($\Delta x$, $\Delta y$ and $\Delta z$) on the accuracy of thickness determination. We assumed that the estimated thickness ($T_{fc}$ and $T_{ac}$) was obtained under the condition that the normal orientation was known. The numerical simulation was performed in the frequency domain exactly in the same manner as described in Sect. 2. Based on thin structure parameters $\tau_{fc}$, $w$, $\tau_{ac}$, $\varphi$, $\theta$, $L_s$, $L_{b+}$ and $L_0$, MR imaging parameters $\Delta x$, $\Delta y$ and $\Delta z$, and Gaussian SD, $\sigma$, $f'(s)$ and $f''(s)$ were obtained according to Eqs. (16) and (17), respectively. Using $f'(s)$ and $f''(s)$, thickness was estimated using Eq. (18). Subsequently, estimated thickness $T_{ac}$ (or $T_{fc}$) was compared with the true thickness $\tau_{ac}$ (or $\tau_{fc}$) to reveal the limits on accuracy. It should be noted that only 1-D computation was necessary for 3-D thickness determination in our numerical simulation. Finally, the plate phantoms were used for validating the theoretical simulation.

The results of simulated thickness measured by the zero crossing method are shown in Fig. 13. For the single thin structure, the measured thickness was an overestimate of the true value for true thickness less than 1.5 mm. For the
two adjacent thin structures, the measured thickness of the $STR_{ac}$ was an underestimate of the true value when the interval $w$ was less than 1.5 mm (except in some cases where the true thickness was 1 mm or less). With an original pixel size of 0.625 mm, image features smaller than 1.5 mm are not accurately represented, so there are measurement errors if either the thickness or the gap is less than 1.5 mm.

Figure 14 shows the average of actually measured thickness obtained from MR images of phantoms with five different $\varphi$ ($\varphi = 0^\circ, 15^\circ, 25^\circ, 35^\circ$, and $45^\circ$) and the fixed $\theta$ ($\theta = 0^\circ$), and the plots of the simulated thickness. The bias between the simulated thickness and the average of actually measured thickness was lower than 0.1 mm, and the SD of actually measured thickness was within 0.1 mm (not shown). A good agreement between the simulated and actually measured thickness is shown in Fig. 14. In the case of $w = 0.5$ mm, thickness of the thin structure was considerably underestimated (except for $\tau_{ac} = 1$ mm). In the case of $w = 1$ mm, thickness of the thin structure was slightly underestimated. In the case of $w = 1.5$ mm, measured thickness values were a good approximation of their true thickness (except for $\tau_{ac} = 1$ mm). From the results of Fig. 14, it can be seen that the effect of $\varphi$ on measured thickness was shown to be sufficiently small in the numerical simulation and MR measurements.

Figure 15 shows the effects of normal orientation $\theta$ and voxel anisotropy $\Delta z$ on measured thickness. Good agreement between the simulated and thickness measured from the MR data of phantoms was observed. The measured thickness was considerably influenced by the latitude angle $\theta$. In the MR data with routine imaging ($\Delta x = 0.625$ mm, $\Delta z = 1.5$ mm, and $\Delta z/\Delta x = 2.4$), the effect of angle $\theta$ on measured thickness was nonmonotonic, i.e., depending on $\tau_{ac}$ measured thickness $T_{ac}$ varied nonmonotonically with the increase of $\theta$.

4.2 Accuracy Analysis of the Two Edge Detection Methods in 2-D

Boundary detection accuracy was determined using the 3 cadaver hip specimens. Here, we investigated accuracy for the radii of $r_1$, $r_2$, $r_3$, $r_4$ as well as for cartilage thickness (see Fig. 9(b)). For an accuracy analysis a so-called gold stan-

![Fig. 13](image_url) Numerical thickness values measured by the zero-crossing method in the 2-D image plane.

![Fig. 14](image_url) The relations between measured thickness and longitude angle $\varphi$. $\Delta x = 0.625$ (mm), $\Delta z = 1.5$ (mm), and $\sigma = (2^{3/2}/2)\Delta y$ ($\Delta y = 0.625$ mm). Graph shows the averages of the actually measured thickness from the MR data of the phantom imaged with the different $\varphi$ while $\theta$ was fixed to $0^\circ$ and the simulated thickness. For phantom measurement, its average ($N = 50$) was indicated by symbols. (a) A single thin structure. (b) Two adjacent acrylic plates with the interval of $w = 0.5$ mm. In the case of $w = 0.5$ mm, the adjacent structure ($STR_{ac}$) had a greater influence on the measured thickness of $STR_{ac}$. (c) Two adjacent acrylic plates with the interval of $w = 1$ mm. In the case of $w = 1$ mm, $STR_{ac}$ had a slight influence on the measured thickness of $STR_{ac}$. (d) Two adjacent acrylic plates with the interval of $w = 1.5$ mm. In the case of $w = 1.5$ mm, measured thickness of $STR_{ac}$ was not influenced by its adjacent structure ($STR_{fc}$). The effect of $\varphi$ on measured thickness was shown to be sufficiently small in the numerical simulation and MR measurements.
Table 2  Accuracy analysis for the radii of $r_1, r_2, r_3, r_4$ as well as for cartilage thickness using the two different methods on the measurement points located at 9 different $\varphi$ and the fixed $\theta (\theta = 0^\circ)$. (all values given in mm).

| Longitude | Anatomic | Femoral Cartilage | Acetabular Cartilage |
|-----------|----------|-------------------|----------------------|
|           | JSW      | $r_1$  | $r_2$ | $r_3$ | $r_4$ | $\sigma$ | $\sigma$ |
| No.1      |          |        |        |        |        |        |        |
| $90^\circ$| 0.47     | 21.82  | 21.91  | 21.76  | 23.60  | 23.66  | 23.17  | 1.78  | 0.625 (mm) | 0.625 (mm) | 2.91  | 2.82  | 2.86  | 2.47  | 2.53  | 2.10  | 2.61  | 2.12  | 2.60  | 2.11  | 2.08  | 2.06  | 1.86  | 1.71  | 1.57  |
| $90^\circ$| 0.51     | 21.61  | 21.52  | 21.50  | 23.67  | 23.65  | 23.23  | 2.06  | 1.73  |
| $90^\circ$| 0.53     | 21.37  | 21.26  | 21.28  | 23.53  | 23.43  | 23.05  | 2.16  | 1.77  |
| $90^\circ$| 0.63     | 21.22  | 21.29  | 21.16  | 23.53  | 23.49  | 23.28  | 2.31  | 2.12  |
| $90^\circ$| 0.66     | 21.05  | 21.12  | 20.93  | 23.52  | 23.65  | 23.03  | 2.47  | 2.10  |
| $90^\circ$| 0.87     | 20.82  | 20.76  | 20.77  | 23.33  | 23.22  | 22.96  | 2.51  | 2.19  |
| $90^\circ$| 1.01     | 20.71  | 20.62  | 20.60  | 23.40  | 23.23  | 23.17  | 2.69  | 2.57  |
| $90^\circ$| 0.93     | 20.53  | 20.46  | 20.43  | 23.34  | 23.33  | 23.06  | 2.81  | 2.63  |
| $90^\circ$| 0.81     | 20.56  | 20.47  | 20.46  | 23.52  | 23.52  | 23.13  | 2.96  | 2.67  |
| No.3      |          |        |        |        |        |        |        |        |        |
| $90^\circ$| 0.53     | 22.22  | 22.19  | 22.17  | 23.79  | 23.80  | 23.39  | 1.57  | 1.22  |
| $90^\circ$| 0.57     | 21.77  | 21.80  | 21.70  | 23.43  | 23.47  | 23.06  | 1.66  | 1.36  |
| $90^\circ$| 0.76     | 21.33  | 21.28  | 21.39  | 23.15  | 23.18  | 23.00  | 1.82  | 1.61  |
| $90^\circ$| 1.02     | 21.11  | 21.08  | 21.17  | 23.16  | 23.20  | 23.12  | 2.05  | 1.95  |
| $90^\circ$| 1.16     | 20.79  | 20.80  | 20.81  | 23.05  | 23.07  | 23.01  | 2.26  | 2.20  |
| $90^\circ$| 1.11     | 20.32  | 20.23  | 20.26  | 22.63  | 22.59  | 22.47  | 2.32  | 2.21  |
| $90^\circ$| 0.79     | 20.51  | 20.55  | 20.46  | 22.88  | 22.97  | 22.52  | 2.37  | 2.06  |
| $90^\circ$| 0.91     | 20.20  | 20.12  | 20.15  | 22.72  | 22.73  | 22.53  | 2.52  | 2.38  |
| $90^\circ$| 0.86     | 19.97  | 20.08  | 19.93  | 22.72  | 22.74  | 22.39  | 2.75  | 2.46  |

1: Anatomic measurement; 2: Proposed method; 3: Zero-crossing method. JSW = Joint space width; FCT = femoral cartilage thickness; ACT = acetabular cartilage thickness. MR imaging was performed through the anatomic coronal slice. For each specimen, anatomic coronal slice and corresponding MR section through the center of the femoral head were analyzed at $10^\circ$ increments (range, $10^\circ$ ~ $90^\circ$) of longitude $\varphi$ along the radial directions from the center of the femoral head, producing 9 measurement points. Anatomic measurements were performed in the coronal plane only.
The proposed model-based method is more accurate for this case. In the same measurement accuracy. Diff

The case for which the methods were significantly different were marked with an asterisk in Fig. 16(a).

Figure 16 shows the mean measurement error and standard error of the measurement error for the zero-crossing method or model-based method - anatomical measurement. Boundary positions were localized on the measurement points at which anatomical hip joint space widths were 0.50 – 0.75 mm and 1.25 – 1.50 mm. An asterisk (*) indicates the two methods differ at the p < 0.05 significance level. (a) Anatomical hip joint space widths being 0.50 – 0.75 mm. (b) Anatomical hip joint space widths being 1.25 – 1.50 mm.

### Table 3: Magnitude of hip joint space width and the femoral and acetabular cartilage thickness on the measurement points located at 9 different θ and the fixed ϕ (ϕ = 0°). (all values given in mm).

| Latitude | Anatomic JSW | ACT | FCT |
|----------|--------------|-----|-----|
|          | 1 | 2 | 3 | 1 | 2 | 3 | 1 | 2 | 3 |
| No.1     |               |    |    |    |    |    |    |    |    |
| 90°      | 0.47          | 2.63 | 2.66 | 1.99 | 1.78 | 1.83 | 1.85 |
| 80°      | 0.51          | 2.57 | 2.72 | 2.13 | 2.06 | 1.95 | 2.03 |
| 70°      | 0.53          | 2.51 | 2.56 | 1.91 | 2.17 | 2.08 | 1.80 |
| 60°      | 0.63          | 2.22 | 2.11 | 1.92 | 2.31 | 2.29 | 1.95 |
| 50°      | 0.66          | 2.11 | 2.19 | 1.88 | 2.47 | 2.52 | 1.76 |
| 40°      | 0.87          | 1.92 | 1.83 | 1.55 | 2.51 | 2.60 | 2.23 |
| 30°      | 1.01          | 1.90 | 1.96 | 1.87 | 2.69 | 2.71 | 2.59 |
| 20°      | 0.95          | 1.83 | 1.77 | 1.73 | 2.81 | 2.71 | 2.67 |
| 10°      | 0.81          | 1.75 | 1.67 | 1.51 | 2.96 | 2.87 | 2.60 |

| No.2     |               |    |    |    |    |    |    |    |    |
| 90°      | 0.53          | 3.31 | 3.36 | 2.35 | 1.57 | 1.56 | 1.91 |
| 80°      | 0.57          | 3.17 | 3.22 | 2.29 | 1.66 | 1.59 | 2.01 |
| 70°      | 0.76          | 2.96 | 3.11 | 2.05 | 1.82 | 1.91 | 1.83 |
| 60°      | 1.02          | 2.82 | 2.91 | 2.32 | 2.05 | 2.02 | 1.87 |
| 50°      | 1.16          | 2.61 | 2.66 | 2.35 | 2.26 | 2.31 | 2.03 |
| 40°      | 1.11          | 2.29 | 2.32 | 1.92 | 2.32 | 2.27 | 2.01 |
| 30°      | 0.79          | 1.98 | 1.95 | 1.83 | 2.37 | 2.31 | 2.05 |
| 20°      | 0.91          | 1.76 | 1.65 | 1.60 | 2.52 | 2.41 | 2.43 |
| 10°      | 0.86          | 1.67 | 1.55 | 1.62 | 2.75 | 2.82 | 2.57 |

| No.3     |               |    |    |    |    |    |    |    |    |
| 90°      | 0.51          | 3.17 | 3.23 | 2.27 | 1.62 | 1.71 | 1.87 |
| 80°      | 0.55          | 3.12 | 3.17 | 2.11 | 1.76 | 1.88 | 2.03 |
| 70°      | 0.79          | 3.06 | 3.17 | 2.13 | 1.90 | 1.81 | 2.00 |
| 60°      | 1.11          | 2.67 | 2.62 | 2.30 | 2.03 | 2.07 | 1.91 |
| 50°      | 1.43          | 2.42 | 2.36 | 2.28 | 2.25 | 2.12 | 2.11 |
| 40°      | 1.57          | 2.21 | 2.32 | 2.23 | 2.47 | 2.56 | 2.35 |
| 30°      | 1.51          | 1.96 | 1.87 | 1.91 | 2.61 | 2.55 | 2.67 |
| 20°      | 1.30          | 1.72 | 1.67 | 1.77 | 2.78 | 2.81 | 2.83 |
| 10°      | 1.21          | 1.62 | 1.63 | 1.55 | 2.91 | 2.80 | 2.87 |

1: Anatomic measurement; 2: Proposed method; 3: Zero-crossing method. JSW = Joint space width; FCT = Femoral cartilage thickness; ACT = Acetabular cartilage thickness. MR imaging plane was perpendicular to the anatomic coronal slice. For each specimen, anatomic coronal slice and corresponding MR section through the center of the femoral head were analyzed at 10° increments (range, 10° – 90°) of latitude angle θ along the radial directions from the center of the femoral head, producing 9 measurement points. Anatomic measurements were performed in the coronal plane only.
Fig. 17  Statistical comparison between accuracy of the proposed and zero-crossing methods for cartilage measurement with reference to SPG measurement. There were 75 measurements over the 2 cadaver hips. (a) Cartilage thickness measured by the zero-crossing method plotted against SPG cartilage thickness. The line of best fit constructed at regression analysis and the corresponding regression equation are shown. Linear regression analysis shows good agreement between cartilage thickness measured by the zero-crossing method and SPG thickness ($r^2 = 0.87$, $p < 0.01$). However, the slope (0.61) and intercept (0.50) of the regression line was significantly different from one and zero, respectively ($p < 0.01$). The $t$-test shows that the difference between the zero-crossing and SPG measurements was statistically significant ($p < 0.005$). (b) Cartilage thickness measured by the new proposed method plotted against SPG cartilage thickness. The line of best fit constructed at regression analysis and the corresponding regression equation are shown. Linear regression analysis yielded $r^2 = 0.99$, the slope and intercept of the regression line being 1.02 and -0.03, respectively. This shows that there was not only a strong linear relationship between the two measurements, but also very good agreement between the values obtained with both methods. The $t$-test shows that the difference between the model-based and SPG measurements was not statistically significant ($p = 0.76$).

4.3 3-D Cartilage Thickness Comparison between Two Different Methods

4.3.1 Anatomic Thickness Measurement as a Gold Standard

3 fresh-frozen cadaver specimens of normal human hip joints were examined. Anatomic thickness measurements were used as a gold standard. We performed the thickness measurements of the acetabular and femoral cartilages on the measurement points located at 9 different $\theta$ and the fixed $\phi$. Starting from the center determined from a circle manually positioned around the radial lines at an angle $\theta$ increment of 10° was used to measure the 3-D cartilage thickness (Table 3). The results show that the new model-based method gave results similar to those presented from anatomic sections, while the zero-crossing method gave large measurement errors relative to anatomic thickness. The $t$-test shows that the two methods were significantly different ($p < 0.005$).

4.3.2 Stereophotogrammetric (SPG) Measurement as a Gold Standard

2 cadaver hip specimens were used to perform the linear regression and correlation analyses for examining the relationships between measurements obtained with the model-based method and SPG measurements, and between measurements obtained with the zero-crossing method and SPG measurements. Figure 17(a) shows a regression relationship ($y = 0.61x + 0.50$) with both the slope and the intercept differing significantly from one and zero, respectively ($p < 0.01$). The $t$-test shows that the difference between both measurements was statistically significant ($p < 0.005$). Figure 17(b) shows a regression relationship ($y = 1.02x - 0.03$) closely approximating the line of identity with neither the slope nor the intercept. This shows that there is not only a strong linear relationship between the two measurements, but also very good agreement between the values obtained with both measurements. The $t$-test shows that the difference between both measurements was not statistically significant ($p = 0.76$).

Figure 18 indicates the cartilage thickness differences over a cadaver hip. We evaluated the accuracy of quantification in cartilage thickness from MRI relative to a “gold standard” SPG measurement. In this figure, the positive difference ranged from 0.03 to 0.57 mm (mean, 0.37 ± 0.28 mm) for the zero-crossing method and ranged from 0.06 to 0.23 mm (mean, 0.13 ± 0.07 mm) for the proposed method; the negative difference ranged from -0.05 to -0.86 mm (mean,
Fig. 18  The color plots indicating the cartilage thickness differences over a cadaver hip. 3-D cartilage thickness distributions were projected onto a plane to illustrate the difference of cartilage thickness. We evaluated the accuracy of quantification in cartilage thickness from MRI relative to a “gold standard” SPG measurement. Cartilage thickness difference maps of the acetabulum, and femoral head with sub-regions outlined: 1: inferior; 2: anterior; 3: superior; 4: posterior. Up: thickness differences between the SPG thickness and the MR thickness using the zero-crossing method. Thickness difference = thickness by using the zero-crossing method – thickness measured by using the SPG method. (a) Acetabulum; (b) femoral head. Down: thickness differences between the SPG thickness and the MR thickness using the new model-based method. Thickness difference = thickness by using the new model-based method – thickness measured by using the SPG method. (c) Acetabulum; (d) femoral head.

-0.42 ± 0.35 mm) for the zero-crossing method and ranged from -0.05 to -0.22 mm (mean, -0.11 ± 0.05 mm) for the proposed method.

4.3.3 Clinical Thickness Measurement

Figure 19 shows the medial view of an example cartilage thickness map for one of the subjects calculated from MR data. The t-test was applied to the measurements made at each discrete region of the patient hips, defined by the longitude and latitude grid. In the region with latitude of 50°–75°, for the thicker cartilage (more than 2.4 mm) the zero-crossing method underestimated the cartilage thickness while for the thinner cartilage (less than 1.5 mm) the zero-crossing method overestimated the cartilage thickness. In such conditions, two methods are significantly different (p<0.001).

15 hips of patients were used for the study. Cartilage thickness maps shown in Fig. 19 revealed the general tendency for cartilage thickness distribution.
5. Discussion and Conclusions

To determine the thickness from MR images, segmentation, that is, boundary detection, of two adjacent thin structures is needed before thickness determination. Segmentation is often complicated by the limited resolution of the image modality at hand. In the existing methods, the boundary detection based on zero-crossings of the second derivatives, or equivalently maxima of the first derivatives, combined with Gaussian blurring is a widely used technique. Using both theoretical simulation and phantom scans, we evaluated the zero-crossing method for determining the boundary location of two adjacent thin structures in the image plane. The results showed that the accuracy of the zero-crossing method was dependent on the gap between two adjacent thin structures (i.e., joint space width) and system spatial resolution. When the gap is less than 1.5 mm, signal intensities of two adjacent thin structures influence each other. The existing methods cannot overcome these limits.

To this end, we describe a new model-based method for estimating the edge locations for the two adjacent thin structures. Two adjacent thin structures separated by a small gap and MR system spatial resolution (i.e., PSF) are modeled. The new method greatly reduces PSF induced bias by incorporating PSF directly into boundary location estimation. The boundary location estimation problem is formulated as a least-square modeling of the actual intensity profile observed in the MR data set with the predicted intensity profile. We have shown that the new approach is more accurate than the zero-crossing method at estimating the boundary locations for two adjacent thin structures in the image plane.

The work described in this paper is related to the technique developed by Chan et al. for measurement of small-vessel from X-ray cine-angiograms [17]. Chan et al. propose a new method that models the scanning process as a linear convolution between a cylinder model and a scanner model. However, because Chan et al. considered the effects of system spatial resolution and object size on vessel measurements of a simple cylinder structure, it is difficult to extend the method to more complex structure (two adjacent thin structures) in MR images. In addition, we present a theoretical analysis for revealing how the JSW can affect the segmentation accuracy of hip cartilages when applying the zero-crossing method.

In conclusion, the zero-crossing method yields considerable biases in boundary detection and thickness measurement of hip cartilage from MR images. The proposed method is more accurate than the zero-crossing method for detecting the hip cartilage boundaries and measuring the hip cartilage thickness. We suggest that the proposed method can be advantageous for measuring the thickness of the articular cartilage in the hip joint.

The current method opens a wide range of applications in experimental studies and clinical investigations on boundary detection and thickness quantification of the complicated thin structures, such as wrist cartilage or brain cortex. In these cases, the adjacent structures influence each other. Thus, our future work will focus on extending the proposed method to the boundary detection and thickness measurement of complicated thin structures from MR or CT images.

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