TECHNICAL REPORT

Role of MRI for detecting micro cracks in teeth

¹Djaudat Idiyatullin, ¹Michael Garwood, ²Laurence Gaalaas and ³Donald R Nixdorf

¹Center for Magnetic Resonance Research and Department of Radiology, University of Minnesota, Minneapolis, MN, USA;

²Department of Diagnostic and Biological Sciences, University of Minnesota, Minneapolis, MN, USA; ³Division of TMD and Orofacial Pain and Department of Neurology, University of Minnesota, Minneapolis, MN, USA

Objectives: To evaluate the limit of tooth crack width visualization by two MRI pulse sequences in comparison with CBCT.

Methods: Two extracted human teeth with known crack locations and dimensions, as determined by reference standard microCT, were selected for experimental imaging. Crack location/dimension and the presence of common dental restorative materials such as amalgam were typical of that found clinically. Experimental imaging consisted of conventional CBCT scans and MRI scans with two pulse sequences including Sweep Imaging with Fourier Transformation (SWIFT) and gradient echo (GRE). CBCT and MR images of extracted teeth were acquired using acquisition parameters identical to those used for *in vivo* imaging. Experimental and reference standard images were registered and the limit of tooth crack visualization was determined.

Results: Collected images indicate that SWIFT could demonstrate cracks with 20- μ m width, which is 10 times narrower than the imaging voxel size. Cracks of this size were not visible in GRE images, even with a short echo time of 2.75 ms. The CBCT images were distorted by artefacts owing to close location of metallic restorations.

Conclusions: The successful visualization of cracks with the SWIFT MRI sequence compared with other clinical modalities suggests that SWIFT MRI can effectively detect microcracks in teeth and therefore may have potential to be a non-invasive method for the *in vivo* detection of cracks in human teeth.

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Introduction

One long-standing problem in dentistry that has great clinical significance is the inability to detect cracks in teeth reliably. A crack in a tooth is a common finding within at least one molar tooth of 70% of randomly selected patients in a subset of dental practices.¹ At present, cracks in teeth are detected by visual findings from inspection, with and without aides such as dyes, magnification and transillumination. These approaches are limited in their ability to determine the extent of the crack and are not able to detect cracks apical to

restorations or within the root structure.² Current X-ray-based techniques, specifically CBCT, have been reported to improve the detection of cracks within the roots of teeth, but systematic reviews suggest that the available research is inconclusive and the method is not reliable.³ The majority of cracked teeth have been restored and imaging these teeth with CBCT produces artefacts that obscure the coronal structures and the cracks as well.⁴ Also, X-ray techniques use ionizing radiation, which is known to be carcinogenic with high or repeated exposure.⁵

MRI has proven itself as an optimal technology for the non-invasive and non-ionizing visualization of soft tissues. The direct imaging of calcified tissues by using echo-based conventional MRI, however, is not possible

Correspondence to: Dr Djaudat Idiyatullin. E-mail: idiat001@umn.edu

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owing to the ultrashort relaxation time of transverse magnetization, which is usually shorter than the minimal echo time (TE) ($TE \approx 1$ ms).

The concept of using MR for imaging in dentistry has become more attractive with the development of methods allowing the direct imaging of densely calcified tissues, such as the dentin and enamel. Three of these methods, including ultrashort TE, free induction decay (FID) projection-based methods like zero TE and Sweep Imaging with Fourier Transformation (SWIFT), were shown to be capable of producing high-quality images of dental tissues *in vivo*.^{6–8}

The purpose of this study was to demonstrate the feasibility of crack detection by MRI technology and compare the results with CBCT.

Methods and materials

Overall approach

Extracted human teeth with known fractures were imaged with CBCT and two experimental MRI protocols using an intraoral coil⁹ (Figure 1b). All acquisition parameters were

identical to those used for *in vivo* MRI and CBCT (Figure 1c) protocols. MicroCT was used as the reference standard to establish crack location and size (Figure 1e). Experimental and reference standard images were registered and the limit of tooth crack visualization was determined by the authors. The contrast-to-noise ratio (CNR) of the visualized small cracks was calculated.

Teeth samples

Two molar teeth selected for imaging were harvested as waste tissue without maintaining any patient-identifying data, thus making their use exempt under current institutional review board protocols. These teeth were selected because they presented a spectrum of crack location, crack dimension and presence of common dental restorative materials (Figure 1a). The teeth were stored in isotonic saline solution, containing 10% formalin as an antimicrobial agent for infection control and to prevent desiccation and secondary fracture (Figure 1d).

X-ray imaging

Three-dimensional (3D) microCT (Metrix®, model XT H 225; Nikon Metrology, Brighton, MI) was obtained

Figure 1 The teeth samples (a), *in vivo* setup for Sweep Imaging with Fourier Transformation (b) and CBCT (c) experiments, vials to fix the teeth (d) and selected orthogonal slices of microCT images (e) with arrows pointing to three natural cracks labelled as c1, c2 and c3.

in one scan using 90 kV, 90 μ A, 708 ms of exposure, 720 projections and four frames per projection. The resolution of the specimens after reconstruction was 7 μ m. The width of the cracks was estimated by using a public domain, Java-based image-processing program developed by Wayne Rasband at the National Institutes of Health (http://imagej.net/Wayne_Rasband).

3D CBCT (17–19 iCAT, Imaging Sciences, Hatfield, PA) was obtained in one scan, with a 60-mm field of view at 37 mA s⁻¹ for 27 s and 120 kV with a nominal resolution of 0.2 mm.

MRI experiments

All MRI experiments were performed in a 4.0-T (human) magnet (Oxford, UK) of 90-cm bore size equipped with an Agilent DirectDrive console (Palo Alto, CA). Radiofrequency transmission and signal reception were performed with a home-built, single-loop, 50-mm-diameter intraoral coil.⁹

In the SWIFT sequence,¹⁰ radiofrequency excitation was performed with an amplitude- and frequency-modulated pulse, commonly called the “hyperbolic secant pulse”, with a stretching factor of 2, a time–bandwidth product of 64, an excitation bandwidth of 100 kHz and flip angle (θ) of 8°. Data were collected in 64 gaps (of 7.4 μ s each) in the RF pulse and after the pulse, 192 samples were acquired without gaps. The repetition time, including the 0.64-ms

pulse length, was 2.6 ms. Data in k -space consisted of 64,000 spokes with termini describing the isotropically distributed points on a sphere. After acquiring a full set of frequency-encoded projections, 3D images are reconstructed with CMRRpack v. 0.45b SWIFT software.¹¹

The gradient-echo (GRE) MRI acquisition used Cartesian k -space sampling with 256 readout points with 10- μ s dwell time and 192 $\times$ 192 phase encodings. Repetition time and TE values were 5.46 ms and 2.75 ms, respectively, and θ was 15°.

The field of view for all MRI experiments was 120 $\times$ 120 $\times$ 120 mm³ and the total acquisition time for each experiment was equal to 3.5 min. All MR images were reconstructed to nominal resolution with a 0.27-mm isotropic voxel size.

MR, CBCT and microCT images were registered and reviewed by two of the authors (DN and DI) to establish the limit of crack visualization on CBCT and MR images. The dimension of fracture at the limit of visualization was established from microCT images. The CNR values of visible small cracks on the SWIFT MR images were determined from magnitude images as: signal-to-noise ratio (SNR) = S/σ and CNR = SNR_{dentin} – SNR_{crack}, where S is the amplitude of target signal, σ is the standard deviation calculated from locations in the image outside the tooth, and SNR_{dentin} and SNR_{crack} are the SNR at dentin and crack, respectively.

Figure 2 An example of *in vivo* Sweep Imaging with Fourier Transformation images of teeth obtained with intraoral coil (a) and optical photo (b).

RESULTS

Figure 2 shows example *in vivo* SWIFT images of teeth obtained with the same intraoral coil and experimental parameters as used in this work for *in vitro* imaging. Image artefacts from the presence of amalgam restorations located on both jaws are notably lacking in MR images.

The width of the cracks in extracted teeth samples decreased from the coronal to apical direction until transitioning to sound-calcified tissue as determined by the reference microCT images, creating a situation where the detectability of cracks depended on width (Figure 1e). On the coronal slice presented in Figure 1e, for example, three cracks, labelled c1, c2 and c3, were

Figure 3 Selected slices of two cracked teeth images of CBCT (a), Sweep Imaging with Fourier Transformation (SWIFT) (b) and gradient echo (GRE) (c).

measured to have 50-, 30- and 40- μ thicknesses, respectively.

Figure 3 presents the comparison of similar slices of CBCT, SWIFT and GRE images. The presence of cracks in the CBCT slices was obscured owing to artefacts originating from the metallic restoration (Figure 3a), a known problem for CBCT. The cracks were highly conspicuous in the SWIFT images, even though their physical thickness is about 10 times less than the width of an image voxel (Figure 3b). Identifiable cracks in SWIFT images were not visible in GRE images obtained under similar experimental conditions (Figure 3c).

Figure 4 shows montages of position-registered similar slices of 3D MR and microCT images. The side-by-side comparison of these images allowed estimation of the smallest size of crack still detectable by MRI. We found the limit of crack detection on SWIFT MR images to be about 20 μ , with CNR ($\text{CNR} = \text{SNR}_{\text{crack}} - \text{SNR}_{\text{dentin}} = 20 - 17$) of about 3 (at Slice 111 for Crack c2 and at Slice 120 for Cracks c1 and c3) (Figure 4b).

Discussion

Images of extracted cracked teeth acquired using parameters identical to *in vivo* MRI and CBCT experiments demonstrated the advantage of using the short T_2 -sensitive SWIFT MRI sequence to detect cracks relative to CBCT. Specifically, SWIFT MRI could detect cracks with about 20- μm width, which is >10 times narrower than the image voxel. Furthermore, unlike CBCT, the MR images are less affected by the close proximity of metallic restorations. Indeed, it is well known that the presence of some materials with high magnetic susceptibility used in dentistry, such as stainless steel and titanium implants, can considerably degrade the diagnostic quality of MR images for adjacent teeth.¹² In our preliminary experience, MR image artefacts are minimal or absent when the metals cause only small perturbations of the magnetic field homogeneity. Hence, amalgam and gold restorations produce far less image artefacts than those observed with CBCT and conventional radiography.

The reason for such differing sensitivities of CBCT and SWIFT MRI could be explained by the differences of contrast mechanisms of these two imaging modalities. The mobile water molecules usually filling the cracks do not significantly attenuate X-rays. Therefore, the signal amplitude in CBCT is linearly dependent on the portion of crack volume creating a negative contrast. Alternatively, in the case of MRI, the presence of water in a crack creates a positive contrast enhancement by two mechanisms: first, the concentration of water inside a crack is at least 5 times higher than that in the dentine;¹³ and second, the transverse relaxation rate ($=1/T_2$) of mobile water in cracks is much smaller than that in motion-restricted dentine pores,¹⁴ which makes the signal from the water located in cracks less blurred and more localized inside of the same voxel.

Figure 4 Montages of selected coronal microCT (a) and registered Sweep Imaging with Fourier Transformation (b) images of a cracked tooth with amalgam filling. The cracks are marked with arrows and arrowhead and labelled (c1, c2 and c3). The insertion on the right top corner shows a 15× magnification of a portion of Crack c2. For comparison, the square box shows the size of voxels (270-μm per side) used in MRI.

Identifiable cracks in SWIFT images are not able to be identified in GRE images obtained under similar experimental conditions, which suggest that the T_2^* of the water located in the cracks is in the submillisecond range. Because water in a crack has

a short T_2^* , but long T_2 , we believe that a conventional two-dimensional spin-echo sequence might also be applied to a specific region of interest in support of short T_2 -sensitive 3D imaging of the full dental structure.

At this time, quantitative data on the sensitivity and specificity of MRI methods for the identification of cracks in teeth are unavailable in the literature. Given our observed difference in the abilities of MRI and CBCT to detect such small cracks using clinically applicable imaging parameters, there seems to be potential for MRI to be able to identify cracks in teeth *in vivo*. Such effective detection would constitute a major step forward because currently there is no reliable method for the *in vivo* detection of small cracks within teeth. A next application of this work would be to assess the reliability of *in vivo* crack detection and to better understand the limits of this technology, in an attempt to develop a non-invasive method for detecting the presence and monitoring the progression of cracks in human teeth. A long-term outcome of this research, when such a tool has been developed, would be to help clinicians determine what crack properties (e.g. width, length, location in the tooth, tooth involved) are most likely to result in unfavourable end points, namely pain and structural failure. From that future research, the assessment of preventative strategies can be explored to obtain the overarching goal of helping people retain their natural dentition for a lifetime.

A limitation of this study is the absence of patient motion, a challenge in clinical dental imaging when high-resolution imaging of small structures is desired. We anticipate that the incorporation of advanced post-

processing techniques, such as adaptive retrospective correction of motion artefacts, will successfully address clinically observed motion artefacts.¹⁵ An additional limitation of this study is the use of a 4.0-T magnet strength, which is not routinely used for clinical imaging. The authors have plans to do a 1.5–4.0-T comparison to better mimic clinical conditions. A final limitation of the study was the inclusion of only two molar teeth. We are currently in the planning stages of a follow-up original research study that will employ a structured analysis of a larger sample of teeth including teeth of different anatomic types.

In conclusion, SWIFT MR images of extracted teeth demonstrate promising visualization of tooth cracks as small as 20 µm, which is 10 times narrower than the image voxel. It is suspected that the high CNR of water inside tooth cracks on SWIFT MR images contributes to visualization beyond the voxel resolution limits of the scan. We believe that SWIFT MRI demonstrates the ability to visualize dental hard tissues unlike conventional MRI and furthermore, unlike CBCT, demonstrates a lack of significant image artefacts associated with nearby dental restorations that prevent visualization of tooth cracks. The authors believe that with additional work, *in vivo* SWIFT MRI of teeth may achieve vastly improved visualization of fractures, a currently problematic area of dental diagnostics.

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