Magnetic Resonance Elastography of the Anal Sphincter

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Introduction

Endoanal ultrasound and MRI can characterize architecture of the external and internal anal sphincters (i.e., its thickness and echogenicity or signal intensity) (1, 2) and reveal anal injury, which is associated with reduced anal pressures in fecal incontinence (FI) (3). However, the relationship between imaging findings and anal pressures is incompletely understood. Large defects may be associated with normal pressures and vice versa. The internal sphincter becomes thicker with aging and is thinned in systemic sclerosis and "primary degeneration" (4-6). This sphincter thinning is attributed, without histological confirmation, to fibrosis and the pathophysiology of thickening is unclear. Anal sphincter stiffness has not been assessed by any technique, including MRE, in humans. The goal of this work is to use magnetic resonance elastography (MRE) to assess the stiffness of the anal sphincters *in vivo*. This is a challenging application since the anal sphincter is not externally accessible and surrounds a narrow lumen.

Materials and Methods

MR elastography was performed on a 1.5-T whole-body GE imager (Signa; GE Medical System, Milwaukee, WI), using an 4 channel torso coil and endorectal MRI Probe (MEDRAD, Inc., Indianola, PA). During imaging, low-amplitude mechanical waves at 120 Hz were generated in the anal sphincter with an acoustic driver device, attached to the endorectal MRI probe, in the rectum. The driver device consists of a pneumatically activated passive drum with a weight attached to the flexible membrane. The passive drum is attached to (3) an active driver source by a flexible tube (10). The gradient echo (GRE) MRE scanning parameters included the following: a pulse repetition time (TR) of 50 ms, an echo time (TE) of 18.3 ms, a flip angle of 30^o, an acquisition matrix of 512 x 256, 3 NEX, a slice thickness of 5 mm, a field of view (FOV) of 22 cm, two pair of interleaved, toggling, motion-encoding gradients acquired for each phase encoding view, and 4 time points equally spaced over the harmonic cycle. Phase difference images were generated from corresponding pairs of acquisitions with reversed motion encoding gradients to remove systematic effects (9). The acquisition time was 5 minutes 20 seconds to acquire a single axial 2D slice sensitized to motion parallel to the long axis of the rectum. Anal sphincter MRE was performed in 6 women: 2 FI, 2 constipation, 2 healthy controls (ages 24-69y).

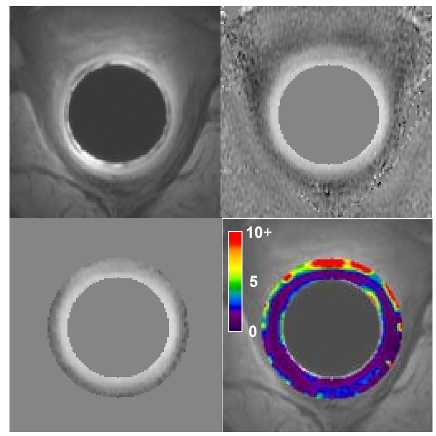

Fig. 1. Sample zoomed (a) magnitude and (b) phase difference image of one volunteer. (c) The smoothed and masked phase difference data. (d) The corresponding stiffness map for this volunteer (in kPa).

Data analysis

Given the rectal geometry and the intra-rectal driving mechanism, the shear waves were assumed to radiate cylindrically outward from the driver, with propagation dominantly in the axial plane. Since the anal sphincter is very thin, the analysis technique chosen was to estimate the spatial gradient of the phase of the propagating wave in radial profiles (11). The internal and external sphincters were manually segmented from the magnitude images. To facilitate analysis, the phase difference data was transformed from polar to rectangular coordinates. Vibration data at the driving frequency was extracted from the four time points, giving a complex number at each pixel representing the amplitude and phase of the harmonic displacement at the driving frequency. Radial shear wave propagation through the sphincter can be documented by a linear change in the phase of this complex number as one moves radially outward. A straight line was fit to the phase of the propagating wave at each angular position (vertical column in the rectangular coordinates). The slope of the straight line fit gives the wave number k and the wave speed is given by $c = 2\pi\lambda/k$. Taking the density of soft tissue to be equal to water, the shear stiffness is given by $\mu = 4\pi f^2/k^2$ (9, 11). Median stiffness values of the internal sphincter of each volunteer were calculated from the pixels within the segmentation mask.

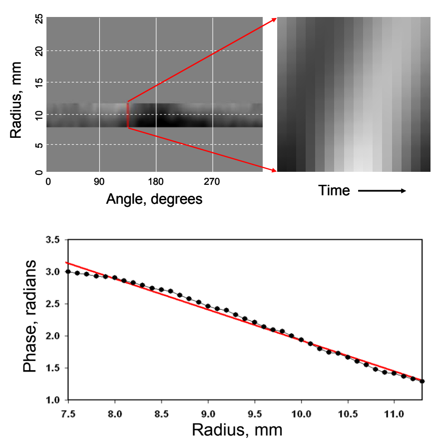

Fig. 2. (Top left) Data similar to fig. 1(c) (from a different volunteer) transformed to rectangular coordinates, and (top right) the data from the indicated column changing over time (16 time points are shown, interpolated from the 4 obtained). (Bottom) The phase of the complex displacement for this column as a function of radius and the corresponding straight line fit.

Results

Fig. 1 shows zoomed magnitude, phase difference, and masked images for one volunteer. The phase difference image depicts displacement in the anal sphincter parallel to the long axis of the rectum at one time point in the harmonic cycle. The average displacement amplitude along the inner radius is 7 microns. Fig. 2 depicts data from another volunteer transformed to rectangular coordinates, and how the displacement in one column of data (one angular position) varies with time, showing a wave propagating upwards in the figure (corresponding to radially outward in the sphincter). The bottom panel shows the phase of the complex displacement as a function of radial position and the corresponding straight line fit. The slope of the straight line fit can be converted to local stiffness as described above. For the volunteer in fig. 1, the median stiffness value was 2.2 kPa. The median value of all volunteers was 3.7 kPa.

Discussion

All 6 volunteers had outward radially propagating shear waves clearly evident over a large fraction of the anal sphincter. Although only 2D data in a single displacement direction was obtained, we believe this is adequate for shear stiffness estimation given the geometry of the rectum and the driver. Median stiffness values in the internal sphincter ranged from 2.2 to 4.3 kPa at 120 Hz. Comparisons to previous studies with similar techniques suggests these values are in the same range as corresponding measurements in the relaxed state for the vastus medialis (i.e., 3.08 ± 0.21 kPa) in children and adults but lower than the biceps brachii (8-34 kPa) and higher than adipose tissue (1.85 ± 0.17 kPa), although these measurements were at varying frequencies (12, 13). Sphincter MRE is challenging since the anal sphincter is a very thin structure, much thinner than shear wavelengths at 120 Hz, and is not externally accessible. The optimal vibration frequency for assessing sphincter stiffness remains to be determined. The value of 120 Hz used here was a compromise between resolution and attenuation based on preliminary data. Preliminary phantom studies of a custom endorectal coil indicate a 4x increase in SNR is possible, which may allow the use of higher frequencies. Further work is also necessary to assess the reproducibility of the technique and establish the range of normal stiffness values and their variation with disease. In conclusion, it is possible to quantify anal sphincter stiffness with MRE.

References

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