Sonoelastography using Compensated Power Doppler

Stephen J. McKenna, Stuart Dickson, Ian W. Ricketts
Department of Applied Computing
University of Dundee
Dundee, Scotland
{stephen,sdickson,ricketts}@computing.dundee.ac.uk

Asif Iqbal, Tim Frank, Alfred Cuschieri
Department of Surgery and Molecular Oncology
University of Dundee
Dundee, Scotland
{a.y.iqbal,t.g.frank,a.cuschieri}@dundee.ac.uk

ABSTRACT
Sonoelastography is the visualisation of elastic properties using ultrasound. It can enable tumours to be detected and localised based on their elasticity when they are less elastic than the surrounding soft tissue. In vibration sonoelastography the target tissues are vibrated while simultaneously recording ultrasound images. A technique for imaging relative elastic properties is proposed that uses a standard ultrasound machine. It combines B-scan and power Doppler signals to produce images of relative vibration amplitude. Preliminary results using simulations and liver phantoms are presented and the potential of the method to highlight areas of differing elasticity within an organ such as the breast is mentioned. The possibility of combining such a method with freehand 3D scanning, enabling B-scan and power Doppler signals to simultaneously populate a voxel array for subsequent visualisation is discussed.

KEY WORDS
Sonoelastography, ultrasound, freehand scanning, power Doppler ultrasound

1 Introduction

The most common clinical method for detecting lumps within tissue, palpation, is highly subjective and dependent on the skill of the practitioner. The method exists because certain pathological conditions, such as malignant tumours, manifest themselves as changes in the tissue's mechanical stiffness. While X-ray imaging is well established for the detection of small, deeply located tumours, X-ray hazards and the desire for better performance have led to a continuing search for alternative techniques. Diagnostic ultrasound is a potential alternative to X-rays, its limitation being that small pathological changes in tissue are difficult to discern on normal ultrasound B-scans. If, however, ultrasonic echo data is collected before and after a slight compression of the tissue, comparisons can be made between normal and pathological areas. This is possible when normal tissues exhibit relatively more movement than stiffer pathological regions. It has been suggestd that benign and malignant tumours can be distinguished by elastography (imaging of elasticity) due to their differing uniformity of elastic properties [14]. Elastography examines the static elastic properties of tissue. Other authors and unpublished results from our own tissue property studies suggest that differences between healthy and pathological tissue are highlighted more clearly using rapidly changing strain. This time-dependent (i.e. viscous) response is analogous to the vibrational frequency response of the tissue. It is likely that a relatively narrow band of vibration frequencies exists for which the response in the tissue is optimum for distinguishing variations in viscoelastic properties. Ultrasound imaging of elastic properties in the presence of vibration is known as sonoelastography. A small, stiff zone will appear as a defined region due to the difference between its motion and that of the surrounding tissue. Ultrasound sonoelastography imaging has been compared to conventional ultrasound imaging for the detection of prostate cancer in vitro, with promising results [18]. Although elastography and sonoelastography are not yet being used in routine clinical practice, these imaging methods have the potential to give comparable spatial resolution to standard grey-scale imaging with enhanced tissue discrimination.

Several authors have reported the use of phase-contrast magnetic resonance imaging (MRI) to visualise the mechanical properties of tissues [2, 5]. Images of tissue subjected to static or time varying displacement are obtained, yielding information on the 3D distribution of elasticity and viscoelasticity respectively. The results from these techniques are very impressive in that small inhomogeneities can be localised. However, the method may never become broadly applicable due to the very high cost of MRI and its lack of portability.

In Doppler ultrasonography, shifts in returning ultrasonic echoes due to the motion of reflecting features are detected. While this is commonly used to highlight blood flow, variations in tissue motion can also be revealed. Ultrasound imaging is particularly attractive as the basic image forming process due to its benign nature and low cost. In addition, the colour Doppler imaging modality offered in modern systems provides motion information when vibration is present. The technique described here, known as Doppler sonoelastography, produces images that can be interpreted in terms of elasticity variations.

Sonoelastography has the potential to enhance the efficacy of screening for common cancers. A primary motivation for conducting the research described here is to lay the foundations for developing a new non-invasive, atrau-

matic and painless system for breast tumour diagnosis and location. The technique is applicable to other common tumour sites such as the prostate. Elasticity imaging may also be used to detect other diseases that cause changes in tissue elasticity (e.g. atheromatous disease). Other applications include the measurement of tissue elasticity for use in tissue modelling studies and to provide data in virtual reality research in relation to surgical/interventional simulators.

In this paper, we investigate the possibility of using the power Doppler imaging mode available as standard on many modern ultrasound machines in combination with co-registered B-scans to perform sonoelastography. Section 2 briefly describes the methods that have previously been proposed for imaging elasticity using ultrasound. In Section 3, a process for estimating the power Doppler signal from tissue under vibration is described and simulated. Simulation results are presented to illustrate the effect of scatterer strength, vibration frequency and vibration amplitude on the power Doppler signal. Section 4 describes preliminary imaging experiments using the theory presented in the previous section. Section 5 describes our freehand 3D scanning system and discusses its applicability to Doppler sonoelastography. Finally, some conclusions and directions for future work are given in Section 6.

2 Elasticity Imaging using Ultrasound

Several techniques for imaging tissue elasticity using ultrasound have been proposed [1, 4, 10, 12, 13, 20]. The reader is refered to Gao *et al.* [7] for a review. Taylor *et al.* [19] classify existing methods as (i) compression elastography (strain imaging), (ii) transient elastography and (iii) vibration sonoelastography. In compression elastography, ultrasound images are compared before and after a compression is applied to the tissue in order to compute a strain map. In transient elastography, a low-frequency transient vibration is applied and the resulting tissue displacement is detected using ultrasound before echoes from tissue boundaries occur. The third class, vibration sonoelastography, which is the topic of this paper, images the vibration patterns resulting when low frequency vibration is applied to the tissue. Vibration propagation within a complex organ cannot be solved analytically. However, small, stiff lesions will tend to result in decreases in vibration amplitude. The extent to which they are contrasted with the surrounding tissue will depend on their size and stiffness and on the frequency of vibration. Losses at high frequencies impose an upper limit on the vibration frequencies that can be used in practice.

Vibration amplitude imaging with low-frequency vibration was first proposed in the late 80's [11]. Tissue motion can be estimated by tracking the two-dimensional image motion of the speckle produced by back-scattering in high frame-rate, real-time ultrasound. Such tracking is often based on template matching methods (e.g. correlation-based motion estimation) although other optical flow algorithms may also be applicable [3]. Speckle tracking can be used to produce images of strain magnitude (e.g. [9]). Finite element methods have been proposed to enable reconstruction of the spatial distribution of Young's modulus [4].

An alternative to speckle tracking is the use of real-time Doppler ultrasound. Doppler techniques measure the component of motion in the direction of ultrasound wave propagation and as such detect axial motion. Doppler ultrasound machines typically use autocorrelation estimators to estimate the mean frequency and the variance of the power spectrum [6]. In flow Doppler, which is used for imaging blood flow for example, the mean frequency is used to estimate the mean velocity. However, under sinusoidal vibration, mean velocity gives no indication of vibration amplitude since oscillation is about a rest position. Taylor *et al.* [19] make use of the relationship between vibration amplitude and flow Doppler variance. They used a scanner specially modified to display the real-time estimate of the variance of the power spectrum. Under reasonable assumptions the standard deviation of the power spectrum is linearly related to vibration amplitude. They were thus able to image the vibration amplitude by measuring this variance.

In contrast, we propose the use of power Doppler imaging in conjunction with co-registered B-scan images to image the vibration amplitude. Power Doppler is available as standard on many ultrasound machines since it is useful for imaging blood flow and has the advantage of good sensitivity to very low velocities.

3 Power Doppler

Power Doppler imaging encodes an estimate of the integrated power Doppler spectrum in pseudo-colour. It cannot be used *directly* to estimate the vibration amplitude because the power Doppler signal depends on the echo strength of the region being imaged as well as its vibration amplitude. The method proposed here uses standard B-scan imaging to compensate the power Doppler signal in order to image the relative vibration amplitudes in the tissue.

3.1 Simulation

Ultrasound scanners tend to use autocorrelation estimators for colour flow and power Doppler imaging. This estimation process was simulated in order to investigate its behaviour at different vibration frequencies and vibration amplitudes. The simulation was similar to that used by Taylor *et al.* [19] to investigate the effect of vibration amplitude on estimates of the *variance* of the Doppler power spectrum as used for flow imaging. Here we investigate estimation of vibration amplitude from the *integrated* power spectrum.

In pulsed Doppler ultrasound, a sequence of ultrasound pulses which together form a packet are used. The number of pulses in a packet is usually user-controlled and here it was set to $N = 16$. These pulses are emitted at intervals of T_p. Each pulse was modelled as a real wavelet

(Equation (1)) where $\sigma = 173ns$ and f_c was the centre frequency.

$$p(t) = \frac{1}{(2\pi\sigma^2)^{\frac{1}{2}}} \exp(-\frac{t^2}{2\sigma^2}) \cos(2\pi f_c t) \quad (1)$$

The round-trip time taken from the ultrasound transducer to a scatterer at distance d_0 and back is $t_{d_0} = 2d_0/c$ where c is the speed of sound. The mean speed of sound in tissue varies from approximately $1446ms^{-1}$ in fat to approximately $1556ms^{-1}$ in spleen, for example. The simulations reported here used $c = 1540ms^{-1}$, an average value used in some scanners [8]. A scatterer under forced vibration was modelled as undergoing sinusoidal motion about a rest position at distance d_0 from the ultrasound transducer with peak vibration amplitude ϵ_m and vibration frequency f_v. Its distance, $d(t)$, from the transducer at time t is therefore given by Equation (2).

$$d(t) = d_0 - \epsilon_m \sin 2\pi f_v t \quad (2)$$

The backscatter, $e(t)$, received from a single pulse at time t is:

$$e(t) = Ap(t - 2d(t)/c) \quad (3)$$

where A is the amplitude of the backscatter and models the echogenicity, the scattering coefficient and attenuation in the tissue. A first-difference FIR filter was applied to the backscattered pulse signals. This simulated a wall filter and attenuated the d.c. component.

Quadrature demodulation was simulated by sampling the N returning, filtered, backscattered pulses in a packet with the first pulse emitted at time $t = 0$. Samples I_n were taken at times $t_I(n) = nT_p + t_{d_0}$ and samples Q_n were taken a quarter of a cycle later at times $t_Q(n) = t_I(n) + 1/(4f_c)$, where $n = 1 \ldots N$. In the experiments described here, $T_p = 1ms$. The average power, R, of the backscattered signal is given by the autocorrelation function at zero lag:

$$R = \frac{1}{N} \sum (I_n^2 + Q_n^2) \quad (4)$$

Note that R is proportional to A^2. Doubling the amplitude of the backscatter, for example, will quadruple the power Doppler signal.

3.2 Simulation Results

Figure 1 shows power, R, plotted against vibration amplitude, ϵ_m for a scatterer vibrated at a frequency of $f_v = 200Hz$ at mean distance $d_0 = 2cm$ from a $7.5MHz$ transducer. Three curves are plotted corresponding to scatterer strengths of $A = 0.1$, $A = 0.2$ and $A = 0.3$. Together these curves illustrate that R increases with A^2. If the power signal can be compensated using an estimate of the scatterer strength these three curves become the same. The power Doppler signal increases monotonically with vibration amplitude over this range ($\leq 1mm$) except at very low power. Therefore an appropriately compensated power Doppler signal could be used to image vibration amplitude in this situation. Note that larger vibration amplitudes would not be properly imaged.

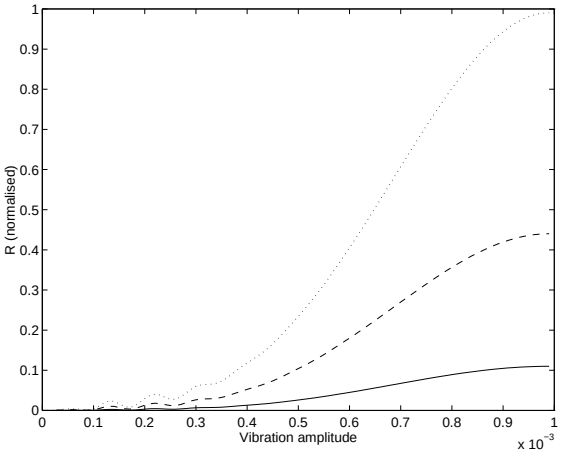

Figure 1. Simulated power Doppler versus vibration amplitude for $A = 0.1$ (solid), $A = 0.2$ (dashed) and $A = 0.3$ (dotted). The power Doppler signal has been normalised. Other parameters were $d_0 = 2cm$, $f_c = 7.5MHz$ and $f_v = 200Hz$

The amplitude and frequency of the vibration source should be selected to be appropriate for the ultrasound transducer used and the depth d_0 of the tissue of interest. This is necessary in order to ensure that the relationship between R and ϵ_m is approximately monotonic and produces a meaningful image for the range of vibration amplitudes induced.

Figures 2 and 3 show three curves generated as in Figure 1 but with the vibration frequency decreased to $40Hz$. Note that the ranges of vibration amplitudes that can now be imaged reliably are different. Figure 2 shows that the range $\epsilon_m = 0$ to $\epsilon_m = 5mm$ could in theory be imaged reasonably well. However, Figure 3 shows that vibration amplitudes in the range $\epsilon_m = 0$ to $\epsilon_m = 250\mu m$ could also be imaged at this depth and frequency.

4 Sonoelastographic Imaging

An Aloka SSD220 ultrasound system with a 7.5MHz linear array probe was used for preliminary imaging experiments. This system has a power Doppler mode as standard.

Several vibration systems have been constructed for experimental use with tissue phantoms. They all consist of a single source using a signal generator and an acoustic speaker. The oscillator incorporated a trigger circuit to allow phase-locking of the image acquisition. The range of vibration frequencies that can be used is limited by loss at higher frequencies. Frequencies need to be selected to enable imaging of the vibration amplitude as described in the previous section. An alternative to sinusoidal vibration is

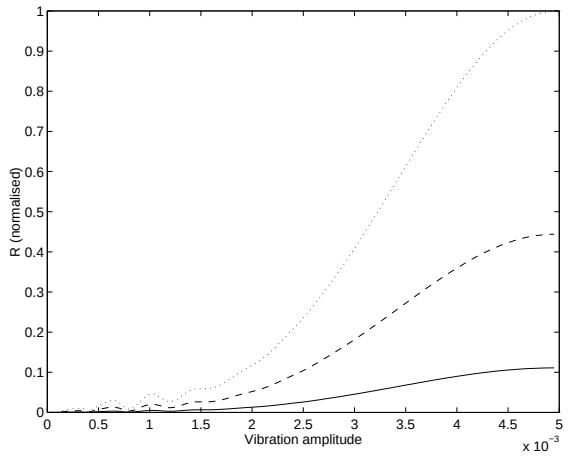

Figure 2. Simulated (nomalised) power Doppler versus vibration amplitude for $A = 0.1$ (solid), $A = 0.2$ (dashed) and $A = 0.3$ (dotted). Parameters were as for Figure 1 with the exception of vibration frequency which was $f_v = 40Hz$.

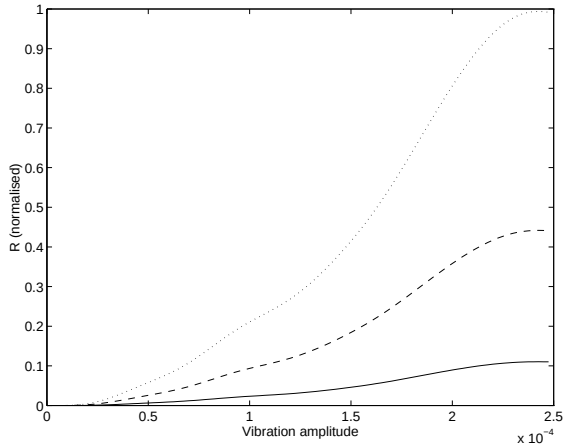

Figure 3. The previous Figure with the abscissa scaled.

to use some form of polychromatic vibration (e.g. square wave vibration). This can help avoid modal patterns and the vibration systems are currently being used to conduct empirical investigation of various forms of vibration over a range of frequencies and amplitudes.

Figure 4 shows co-registered B-scan and power Doppler images of a slice through a commercially available synthetic breast phantom (supplied by Computing Imaging Reference Systems, Inc.). Vibration was applied at the base in the Figures and the transducer was at the top. The frequency of the applied vibration was $30Hz$. The images were obtained by keeping the ultrasound probe clamped in a fixed position while switching from B-scan mode to power Doppler mode. A fluid filled cyst can be seen at the lower right and a stiff lesion at the lower left. The cyst appears as a void in the B-scan and therefore also as a void in the power Doppler image. Since there is no significant backscattered signal from this region, there is no power Doppler signal. The lesion at the lower left appears contrasted in the B-scan indicating that it scatters with greater amplitude. It also appears contrasted in the power Doppler signal and the extent of this contrast is due to both the increased scatter and the increased stiffness.

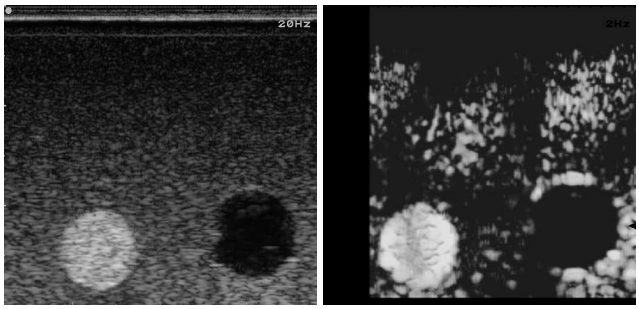

Figure 4. Mean B-scan (left) and power Doppler (right) images of the same slice of a commercial breast phantom with an applied vibration.

Figure 5 shows co-registered B-scan, power Doppler and sonoelastographic images. The sonoelastographic image was obtained by compensating the power Doppler data with squared B-scan data. The square root of the resulting pixel data is shown for better visualisation. The images are of a liver phantom with a simulated tumour in the centre. This tumour appears hyperechoic in the B-scan but is not apparent in the power Doppler signal as a result, despite its lower elasticity. However, it appears as a void in sonoelastographic image indicating that it is stiffer than the surrounding tissue.

5 Freehand 3D Sonoelastography

Standard two-dimensional ultrasound scanners can be used to provide three-dimensional images if the 3D position and orientation of the transducer is known for each of the 2D images recorded. Free-hand 3D scanning can be used to construct a 3D volume (voxel array) while the physician performs an examination in a normal manner. Visualisation software can then provide an operator with the ability to explore the 3D volume using techniques such as any-slice imaging, rendering with transparency and 3D viewpoint control.

5.1 Calibration and Visualisation

A freehand scanning system has been constructed in which the ultrasound probe's 3D position and orientation are measured using a Polhemus Fastrak electromagnetic sensor with an angular and positional accuracy of 0.5° and $0.5mm$ respectively. The Polhemus device is portable and accurate

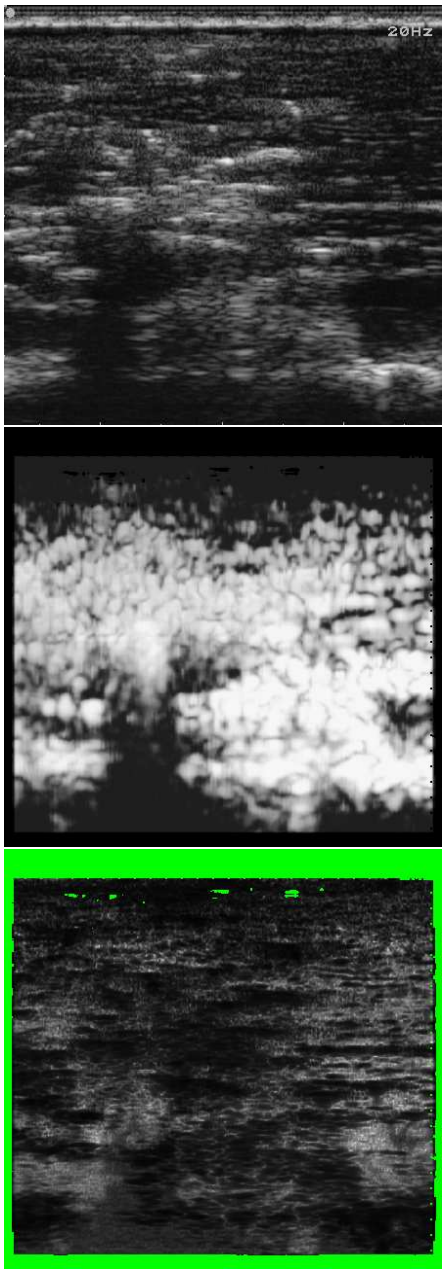

Figure 5. Top: B-Scan, Middle: Power Doppler, Bottom: Sonoelastographic image. The sonoelastographic image was obtained by compensating the Dopper image with squared B-scan data and then taking the square root of each pixel to improve visualisation.

as long as reasonable precautions are taken to avoid electromagnetic interference. Once attached to the ultrasound probe, the Polhemus sensing system is spatially calibrated with respect to the B-scans using an object of known geometry. This spatial calibration was performed using the *StradX* software and a phantom constructed using a similar design to that described by Prager *et al.* [15]. This phantom is shown in Figure 6. The measurements from the Polhemus sensor and the ultrasound images captured were

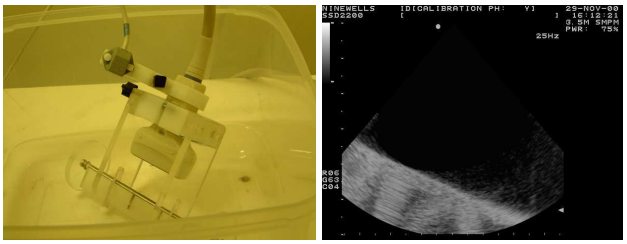

Figure 6. Left: the calibration phantom with the probe and Polhemus sensor attached. Right: a calibration image

temporally aligned using a trigger signal supplied via a footswitch. This system can be used to reconstruct 3D volumes from freehand scans and is being used to investigate the possibility of 3D sonoelastography based on simultaneously acquired B-scan and power Doppler data. High quality 3D reconstruction requires accurate calibration of the free-hand scanning device, accurate registration and data fusion. Image-based registration and fusion can improve the quality of the reconstruction and reduce speckle noise, shadowing and signal dropout [16, 17].

5.2 Current Research Issues for 3D Doppler Sonoelastography

Doppler imaging only measures the component of velocity in the direction of wave propagation i.e. only axial motion can be detected. However, freehand scanning can be used to register and fuse Doppler signals from different transducer orientations.

The process of forming a power Doppler image is relatively slow since multi-pulse packets are used for estimation. This means that the probe can move significantly during acquisition of a single image. Probe position and orientation measurements therefore need to be interpolated over time so that columns in the 2D images obtained from the linear array probe can be aligned appropriately in 3D. The frame-rate for power Doppler imaging is often increased by reducing the area over which Doppler signals are estimated, the remaining area being imaged in B-scan mode. During such a scan, the B-scan and Doppler data can be recorded and registered in two separate 3D voxel arrays. Fusion of these B-scan and Doppler volumes then compensates the power Doppler data to produce 3D sonoelastographic volume data. Each voxel in this volume could be assigned an uncertainty value based on the amount of evidence available from the 2D scans and the extent of the interpolation used. This uncertainty data could in turn provide useful information for subsequent data fusion and provide feedback to the physician performing the scan. Three-dimensional sonoelastographic imaging has the potential application of measurement of the volume distribution of tissue elastic properties. This data is required by researchers developing mathematical models of tissue and, in particular, for use in electronic tissue representation in surgical simulators.

6 Conclusions

A method for vibration sonoelastography has been proposed based on compensating power Doppler images with B-scan image data. Simulations were presented to explore the feasibility of the method and the sensitivity to vibration amplitude and frequency. An experimental imaging system was described and the potential benefits of its extension to freehand 3D scanning systems were outlined.

Future work could extend the simulation using finite element shear wave vibration modelling combined with models of ultrasound propagation. As demonstrated in the simulations presented here, sonoelastographic scanning requires the determination of optimal frequency ranges for the externally applied vibration. Modal patterns (standing waves) can occur, especially for regular shapes, and these can make image interpretation difficult [19]. The use of polychromatic vibration should be explored to alleviate such artefacts. We are also studying Doppler imaging in the non-power mode. Here the image acquisition times are shorter and allow us to synchronise the image to an adjustable phase position within one vibration cycle. This is revealing information about vibration mode structures and will help to develop methods for avoiding them.

An objective of this research is to improve detection of subclinical breast cancer by breast sonoelastography. The sonoelastographic imaging modality is painless, risk free and uses equipment that is relatively cheap and portable.

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