A MR Compatible Mechatronic System to Facilitate Magic Angle Experiments *in Vivo*

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Abstract. When imaging tendons and cartilage in a MRI scanner, an increase in signal intensity is observed when they are oriented at 55 degrees with respect to B_0 (the “magic angle”). There is a clear clinical importance for considering this effect as part of the diagnosis of orthopaedic and other injury. Experimental studies of this phenomenon have been made harder by practical difficulties of tissue positioning and orientation in the confined environment of cylindrical scanners. An MRI compatible mechatronic system has been developed to position a variety of limbs inside the field of view of the scanner, to be used as a diagnostic and research tool. It is actuated with a novel pneumatic motor comprised of a heavily geared down air turbine, and is controlled in a closed loop using standard optical encoders. MR compatibility is demonstrated as well as the results of preliminary trials used to image the Achilles tendon of human volunteers at different orientations. A 4 to 13 fold increase in signal at the tendon is observed at the magic angle.

Keywords: MRI, magic angle, MR compatible robotics, MRI compatibility, medical robotics.

1 Introduction

With conventional MR sequences, normal tendons exhibit very little or no signal intensity due to being comprised predominantly of dense collagen with a highly ordered structure [1, 2]. As a result of this structure, dipole-dipole interactions between nuclei are greatly enhanced which leads to a rapid loss of signal directly after excitation displaying short T2 times. Therefore, very little or no signal is detectable with the Echo Time (TE) available on most clinical systems [2].

As indicated in [1] the dipolar interaction between two nuclei is proportional to a factor of $3\cos^2\theta - 1$ where θ is the angle between the direction along which the spins

are coupled in the structure in which they are held (in this case the tendon), and the direction of the main magnetic field B_0 . This dipolar interaction disappears when the above factor becomes zero, which is satisfied for $\theta = 54.74^\circ$. This value is generally approximated to 55 degrees and termed the “*magic angle*” [1], and the phenomenon of increased signal intensity at the magic angle has been coined the “*magic angle effect*” [3]. It is therefore important to consider the orientation of tissue with respect to the main field when observing increased signal intensity in tendons. Recent studies have demonstrated the clinical and functional importance of exploiting this dipolar coupling effect in cartilage and tendons as part of the evaluation of orthopaedic and other injury [2, 4-6]. To those unfamiliar with this effect, these signal variations may indicate disease, rather than an imaging artifact [7].

The work described in the body of literature documenting the magic angle phenomenon was made very much harder by the practical difficulties of placing a tissue at a specific angle to B_0 in the confined environment of a closed bore MRI system. An actuated system introduced inside the scanner bore which can move the anatomy to a desired orientation would make the process quicker, more accurate and reliable and would allow the position of the tissue to be modified remotely without patient repositioning or scanner bed movement. These advantages would justify the increased cost of such a system with respects to inflexible manual jigs and fixtures. This paper describes the development of a new MRI compatible system designed to assist magic angle related experiments in vivo, as a research and diagnostic tool. The intense restrictions imposed by the MRI environment on mechatronic systems and the high forces required to position a limb encouraged the development of a novel pneumatic rotary motor. The system has been shown to work reliably and the results of pre-clinical trials with human volunteers are presented, showing the increase in signal intensity in the Achilles tendon at and around the magic angle.

2 Description of Limb Positioning System

2.1 System Specifications

The objective of this work is to develop a system which is capable of positioning a variety of target tissues relative to the main field inside a MRI scanner bore. The following specifications were set: (i) the system will be located inside the imaging volume of a closed bore 1.5T or 3T MRI scanner, and should be independent of the scanner make and model; (ii) due to spatial constraints it will be able to position legs from the knee down and arms from the elbow down; (iii) its range of motion should allow orientation change by at least 60 degrees with respect to the main field B_0 while still ensuring patient comfort; (iv) the system should be controlled remotely via an intuitive graphical user interface; (v) it should not be affected by the fields from the scanner and must not deteriorate the MR images.

2.2 Actuation Method

MRI imposes severe restrictions on any mechatronic device introduced in the MR environment [8]. The magnetic fields used by the scanner require the selection of suitable materials, actuators and sensors [9, 10]. Piezoceramic motors have been the most

common substitute for regular DC motors, and are used in a number of systems [11–13]. The high forces required to move human limbs (especially the leg) eliminated the use of piezoceramic actuators in this application. A novel pneumatic air motor system was developed, where air is introduced through a nozzle onto the blades of a turbine rotor, as shown in Fig. 1(a). The output of the rotor is transmitted through a gear train (ratio 4616:1) to produce an adequate speed and torque at the output shaft. The gear-box contains plastic gears (Delrin[®]), aluminium shafts and plastic bearings with glass balls. Both the turbine rotor and the gear box casing (Fig. 1b and c) are manufactured using an epoxy-resin rapid prototyping technique. The operating pressure was chosen to be 1 bar for safety. The motor presents a maximum speed of 6rpm and can output a torque of up to 0.35Nm (using a flow rate of 40 litres/min). The torque is enough to move a subject's limbs as long as he/she is cooperative.

Fig. 1. (a) CAD representation of the operating principle of the pneumatic motor; a flow of air at 1 bar is introduced through a nozzle to move a turbine rotor. Two inlets allow bidirectional motion. (b) and (c) Photographs of turbine, rotor and gear box: 75x75x50mm.

2.3 System Hardware

The system consists of a 1 DOF platform as can be seen in Fig 2(a). The patient's leg is strapped to the platform frame, which is then moved by the air motor via a rack and pinion through the linkage. The slider rail is made of aluminium, and the rest of the hardware is manufactured using Ertalon[®] and Delrin[®], both MR compatible engineering plastics. The base plate is fixed to the scanner bed in one of a number of different initial positions (depending on the size of the patient). The platform can be positioned vertically or horizontally depending on the limb which will be oriented in the scanner. A surface mount optical encoder shown in Fig 2(b) is used to detect the position of the platform (Agilent Technologies AEDR-8000 series). Cables from the encoder are directed into a shielded aluminium box (which acts as a Faraday cage), located inside the scanner room 2m away from the entrance to the bore. Circuits inside the box (powered by alkaline batteries) convert the encoder electrical pulses to fibre optic

signals, which are transmitted to the control room through the wave guides to circuits which transform the optic signals back into electrical pulses. A motion controller (National Instruments CompactRio) then decodes the pulses into position, giving the angle of the platform. The controller runs a proportional-integral-derivative (PID) position control algorithm which controls the air supply to the air turbine by pulse-width-modulating two solenoid valves. The compressed air is supplied by a portable silent compressor in the scanner control room. The position of the platform is specified from a graphical user interface running on a PC inside the control room, developed in LabView (National Instruments), which communicates with the motion controller. A block diagram of the system is presented in Fig 2(c).

Fig. 2. (a) A photograph of the one DOF MRI compatible manipulator and its parts, designed for limb positioning inside a closed bore scanner. (b) Surface mount optical encoder on its PCB and MR images showing 20mm maximum artifact produced and the reference object as specified by the ASTM protocol F2119. (c) Block diagram of limb positioning system. The manipulator is located inside the scanner bore with the encoder electronics inside the room. The rest of the system is inside the control room.

To evaluate the accuracy of the system, the position indicated by the encoders was compared to that of an external mechanical tracking arm attached to the platform. It was always positioned to within an accuracy of ± 0.5 degrees. Once the limb is located at a desirable position, the image is acquired. As the motor is non-backdrivable, that is, its shaft cannot be moved by an external force, the leg is maintained still while scanning is taking place, not requiring a braking mechanism.

3 MR Compatibility Tests

As the manipulator contains no ferromagnetic or highly paramagnetic materials it will not suffer any forces or torques when introduced into the scanner room. To evaluate MR compatibility two tests were performed: (i) to quantify any image artifacts produced by the system when introduced into the scanner bore, and (ii) to measure the impact the system had on the Signal to Noise Ratio (SNR) of the images. The optical encoders were evaluated to measure any artifacts produced, under the protocol described in the ASTM standard F2119 [14]. The encoder gave a maximum artifact of 20mm as can be seen in Fig. 2(b). This size indicates the minimum distance at which this element must be separated from the region of interest of the MR image.

To evaluate the impact of the limb positioning system on the SNR of the MR scans the following images were taken: (i) a control scan of a bottled cylindrical phantom (with a CuSo₄ solution) in the magnet isocentre, (ii) the phantom placed next to the manipulator when powered but not actuated, and (iii) the phantom with the manipulator powered and actuated. In each case the SNR was calculated, using the definition given by the following formula [11]:

$$SNR = \frac{P_{centre}}{SD_{noise}} \quad , \tag{1}$$

where P_{centre} is the mean signal in the 40x40 pixel region in the centre of the phantom, and SD_{noise} the standard deviation of the signal in the 40x40 pixel region at the bottom right hand corner of the image. The tests were performed on a 1.5T Siemens Magnetom Vision using a Turbo Spin Echo (TSE) sequence (TR=2000ms, TE=120ms, 5mm slice thickness, 0.5 distance factor, 256 matrix, FOV=300mm, BW=130Hz/px) and a Gradient Echo (GE) sequence Fast Low Angle Shot (FLASH) (TR=22ms, TE=10ms, Flip Angle=80°, 5mm slice thickness, 0.5 distance factor, FOV=300mm, 256 matrix, BW=130Hz/px).

In Table 1, the values of the SNR for each sequence are presented along with the variation of SNR with respect to the phantom alone. The results show that the impact of the system on the SNR of the images is virtually negligible, and can therefore be considered fully MR-compatible under the reported conditions.

Table 1. Results of SNR measurement test with the system powered and actuated when compared with only a phantom in the scanner

Test	SNR TSE	Variation (%)	SNR GE	Variation (%)
Phantom alone	81.5		25.4	
System Powered	80.1	1.71	25.9	-1.04
System Actuated	81	0.6	24.5	3.54

4 Preliminary Trials with Human Volunteers

Tests were performed on three human volunteers (on the Achilles tendon of the right leg). This tendon was chosen due to its size and ease of location on the body and the

MR images. Initial trials were undertaken in a 1.5T Siemens Magnetom Vision scanner. The unit was mounted vertically in the scanner and left of the centre of the machine (see Fig. 3). The subject was positioned in the scanner bore in the left lateral decubitus position. The right leg was strapped to the platform and a flexible surface coil was wrapped around the Achilles tendon to improve imaging.

The desirable image pattern is one in which the plane of the images is constant relative to the target tissue. This means that, in the case of the Achilles tendon, the initial images (when the tendon is approximately parallel to the main field) are transverse, but, as the angle between tendon and field increases, become increasingly nearer sagittal, and the main images of interest become increasingly further from the centre of the machine. In order to scan the same cross sections on the leg as it is oriented at different angles, a passive fiducial marker made from Spenco[®] [15] was fixed to the subject's leg at the beginning of the session. This served as a reference location on the leg, from which the planes of the tendon to be imaged were defined. The sequence used in the study was a Spin Echo (TR=300ms, TE=20ms, slice thickness 5mm, 5 slices, FOV=250mm, 256 matrix, space between slices 2.5mm). With the leg in place, the platform was actuated in five degree increments starting at zero and images taken at each angle. Due to the pronounced increase in signal after an angle of 45 degrees, an image was taken every two degrees from 45 to 61 degrees.

Fig. 3. The subject's right leg is strapped to the platform and a surface coil is wrapped around the Achilles tendon before being introduced into the scanner bore for imaging

In each case the system could position the lower leg at the desired angle. In Fig. 4(a) and (b) the magic angle effect can clearly be seen on the two MR images. The first displays a transverse scan of the tendon aligned with the main field and displaying virtually no signal and the second shows the tendon at the magic angle with its corresponding increase in signal. Fig. 4(c) presents a graph displaying the average normalised signal of the Achilles tendon (taken over three contiguous slices) for each human volunteer. To compensate for the fact that the sensitivity to magnetisation of the surface coil changes as the platform is rotated, the signal at the tendon is normalised at each orientation, by dividing it with the signal at the tibia. In this way the increase in normalised signal obtained is solely due to the magic angle effect. The graph shows a small signal in the tendon up to an angle of approximately 45°, after which there is a sharp increase in signal that peaks at 55° and then rapidly reduces again. At 55° the signal intensity varies between subjects, but ranges from a 4 to 13 times

Fig. 4. (a) Achilles tendon shown at 0° with respect to the main field and (b) at 55°. There is a clear increase in signal intensity in the Achilles tendon. (c) The graph shows the average values of normalised signal for each subject. The signal in the tendon peaks when the dipole interaction is zero, showing a 4 to 14 fold increase in SNR.

increase than when aligned with B_0 . The graph also shows the dipole-dipole interaction: the signal peaks when this interaction is zero.

5 Discussion and Conclusions

This paper has described an MRI compatible mechatronic system actuated with a novel pneumatic motor for positioning of a subject's limb inside the field of view of a MRI scanner. It is envisaged to be used as a research and diagnostic tool to aid orthopaedic diagnosis and other magic angle related studies independent of the make and model of the MRI scanner. The conditions under which MRI compatibility is assured have been indicated and preliminary trials with human volunteers prove its functionality showing a 4 to 13 fold increase in signal at the tendon. However the system presents some limitations under its present implementation due to its solitary degree of freedom. Current development includes adding additional degrees of freedom to overcome these shortcomings.

References

- [1] Erickson, S.J., Prost, R.W., Timins, M.E.: The "magic angle" effect: background physics and clinical relevance. *Radiology* 188, 23–25 (1993)
- [2] Marshall, H., Howarth, C., Larkman, D.J., Herlihy, A.H., Oatridge, A., Bydder, G.M.: Contrast-enhanced magic-angle MR imaging of the Achilles tendon. *AJR Am. J. Roentgenol.* 179, 187–192 (2002)
- [3] Xia, Y.: Magic-angle effect in magnetic resonance imaging of articular cartilage: a review. *Investigative radiology* 35, 602–621 (2000)
- [4] Oatridge, A., Herlihy, A.H., Thomas, R.W., Wallace, A.L., Curati, W.L., Hajnal, J.V., Bydder, G.M.: Magnetic resonance: magic angle imaging of the Achilles tendon. *The Lancet* 358, 1610–1611 (2001)
- [5] Chappell, K.E., Robson, M.D., Stonebridge-Foster, A., Glover, A., Allsop, J.M., Williams, A.D., Herlihy, A.H., Moss, J., Gishen, P., Bydder, G.M.: Magic angle effects in MR neurography. *AJNR Am. J. Neuroradiol.* 25, 431–440 (2004)
- [6] Fullerton, G.D., Cameron, I.L., Ord, V.A.: Orientation of tendons in the magnetic field and its effect on T2 relaxation times. *Radiology* 155, 433–435 (1985)
- [7] Bydder, M., Rahal, A., Fullerton, G.D., Bydder, G.M.: The magic angle effect: a source of artifact, determinant of image contrast, and technique for imaging. *J. Magn. Reson. Imaging* 25, 290–300 (2007)
- [8] FDA, A Primer on Medical Device Interactions with Magnetic Resonance Imaging Systems, World Wide Web: <http://www.fda.gov/cdrh/ode/primerf6.html>
- [9] Elhawary, H., Zivanovic, A., Davies, B., Lamperth, M.: A Review of Magnetic Resonance Imaging Compatible Manipulators in Surgery. In: *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 220, pp. 413–424 (2006)
- [10] Roger, G., Yamamoto, A., Chapuis, D., Dovat, L., Bleuler, H., Burdet, E.: Actuation methods for applications in MR environments. *Concepts in Magnetic Resonance Part B: Magnetic Resonance Engineering* 29B, 191–209 (2006)
- [11] Chinzei, K., Hata, N., Jolesz, F.A., Kikinis, R.: Surgical Assist Robot for the Active Navigation in the Intraoperative MRI: Hardware Design Issues. In: presented at Proc. 2000 IEEE/RSJ International Conf. Intelligent Robots and Systems, Maui, HI, USA (2000)
- [12] Larson, B.T., Erdman, A.G., Tsekos, N.V., Yacoub, E., Tsekos, P.V., Koutlas, I.G.: Design of an MRI-Compatible Robotic Stereotactic Device for Minimally Invasive Interventions in the Breast. *Journal of Biomechanical Engineering - Transactions of the ASME* 126, 458–465 (2004)
- [13] Tsekos, N.V., Ozcan, A., Christoforou, E.: A prototype manipulator for magnetic resonance-guided interventions inside standard cylindrical magnetic resonance imaging scanners. *J. Biomech. Eng.* 127, 972–980 (2005)
- [14] ASTM, Standard Test Method for Evaluation of MR Image Artifacts from Passive Implants, Designation F2110-01 (2001)
- [15] Spenco Healthcare International Limited, World Wide Web: <http://www.spenco-healthcare.co.uk/>