

A Novel MRI Compatible Soft Tissue Indentor and Fibre Bragg Grating Force Sensor

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Abstract

MRI is an ideal method for non-invasive soft tissue mechanical properties investigation. This requires mechanical excitation of the body's tissues and measurement of the corresponding boundary conditions such as soft tissue deformation inside the MRI environment. However, this is technically difficult since load application and measurement of boundary conditions requires MRI compatible actuators and sensors. This paper describes a novel MRI compatible computer controlled soft tissue indentor and optical Fibre Bragg Grating (FBG) force sensor. The high acquisition rate (100 Hz) force sensor was calibrated for forces up to 15 N and demonstrated a maximum error of 0.043 N. Performance and MRI compatibility of the devices was verified using indentation tests on a silicone gel phantom and the upper arm of a volunteer. The computer controlled indentor provided a highly repeatable tissue deformation. Since the indentor and force sensor are composed of non-ferromagnetic materials, they are MRI compatible and no artefacts or temporal SNR reductions were observed. In a phantom study the mean and standard deviation of the temporal SNR levels without the indentor present were 500.18 and 207.08 respectively. With the indentor present the mean and standard deviation were 501.95 and 200.45 respectively. This computer controlled MRI compatible soft tissue indentation system with an integrated force sensor has a broad range of applications and will be used in the future for the non-invasive analysis of the mechanical properties of skeletal muscle tissue.

Key words: MRI, soft tissue, biomechanics, MRI compatible, actuator, force sensor, Fibre Bragg Grating.

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1. Introduction

Magnetic Resonance Imaging (MRI) is an ideal modality for the non-invasive analysis of soft tissue biomechanics as it provides excellent soft tissue contrast without exposing subjects to ionizing radiation. In addition it allows for the measurement of various biomechanical boundary conditions required for inverse analysis of tissue properties, such as 3D tissue geometry (segmentable from anatomical MRI), 3D architecture (e.g. based on diffusion tensor MRI [1]) and accurate 3D soft tissue deformation measurement (e.g. based on tagged MRI [2, 3]). The non-invasive investigation of soft tissue mechanical properties allows for the validation of detailed constitutive laws enabling derivation of *in-vivo* tissue loading conditions and prediction of injury. Hence it has a wide array of applications including impact biomechanics [4], rehabilitation engineering [5] and surgical simulation [6].

However the MRI based investigation of the mechanical properties of soft tissue often requires an MRI compatible soft tissue loading system consisting of actuators (to mechanically palpate/excite the tissue) and sensor devices (to measure the applied load). Designing such devices to be safe within the MRI environment and compatible with the imaging is non-trivial [7]. Detailed evaluation of constitutive properties (including viscoelasticity) requires dynamic load application capabilities and, since imaging may require motion repetitions, displacement or force control and control of timing with respect to imaging. This paper describes a novel computer controlled MRI compatible soft tissue indentation system and force sensor enabling the MRI based analysis of the non-linear (visco)elastic and anisotropic behaviour of skeletal muscle tissue. The current focus is non-painful and non-damaging indentation of the biceps region of the upper arm using a 45 mm diameter, which places a load constraint in the order of 15 N. Both quasi-static (load is held static during measurement of boundary conditions) and dynamic applications are of interest (non-painful indentation speeds whereby boundary conditions are continuously sampled as the tissue deforms).

Actuator and sensor devices which can safely be used within the MRI environment (are MRI conditional or MRI safe) are relevant to many research areas. As such a large array of MRI compatible actuators [8] (review paper) and sensor systems [9] (review paper) have been proposed. Applications include: MRI robotics (e.g. [10, 11]), MRI guided surgical interventions (e.g. [12-14]), MRI based catheterisation (e.g. [15]), functional MRI (e.g. [16, 17]) and the study of pressure ulcer development (e.g. [18-23]). In addition, as is relevant here, they have been employed for non-invasive investigation of soft tissue biomechanics (e.g. [2, 18, 22-26]).

Two main groups of actuator and sensor devices suitable for the MRI environment can be distinguished: 1) *Devices employing electric principles and/or ferromagnetic components in the MRI room and/or close to the imaging region* and 2) *Devices which are intrinsically MRI compatible since they employ non-conducting and non-ferromagnetic materials and sensor signal transmission occurs using magnetically inert media within the MRI room*. The latter group is the focus here since these do not significantly affect image quality and sensor performance. In addition they do not require shielding or anchoring and can safely be used in close proximity to the test subject. Therefore the applicability,

safety and compatibility with MRI are more easily established for these devices which simplifies evaluation and approval (e.g. ethical) for a research setting.

A variety of soft tissue loading systems that can be used in the MRI environment have been proposed in the literature (e.g. [18, 22-27]), however their design is often tailored to suit their particular application (e.g. ranging from rat lower limb [18], to human footpad [25] and pigs buttock [23]) which, to date, has not included indentation of the upper arm. Despite this, a brief discussion of current soft tissue loading systems for the MRI environment is presented here. Most current soft tissue loading systems do not have dynamic displacement or force control capabilities during imaging (e.g. [22, 24, 25]) and have so far been limited to quasi-static applications (e.g. [18, 22-26]). In some cases the load can only be applied prior to imaging and cannot be altered during scanning (e.g. [22, 24]) which does not enable dynamic analysis and MRI based deformation measurement techniques such as SPAMM (Spatial Modulation of the Magnetisation) (e.g. [2, 3]) tagged MRI. Hence some researchers have employed contour tracking from matched anatomical MRI scans prior to and after loading (e.g. [24]) and registration techniques (e.g. [26]) or finite element analysis (FEA) for estimation of deformation instead (e.g. [21, 22]). However these measures are more limited than full 3D deformation measurement based on SPAMM tagged MRI [3]. In addition the studies employing soft tissue loading devices have limited their analysis to the evaluation of isotropic hyperelastic constitutive models (e.g. [18, 21, 22, 24, 26, 28, 29]). Evaluation of more detailed constitutive models incorporating non-linearity, viscoelasticity and anisotropy require more detailed boundary condition measurement which is the focus of the current study.

Recently more advanced soft tissue loading devices have been proposed for the MRI environment such as Fu et al. 2011 [27] (a hydraulic actuator for lower leg indentation) and Solis et al. 2012 [23] (pigs buttock compression using a flat plate connected by a ~3 m rod (which passes through a wall) to a servo motor and force transducer outside the MRI room). Although both these systems feature dynamic displacement control capabilities and motion triggering toward the imaging, the latter can only operate approximately parallel to the MRI bore axis through the particular hole in the wall and for the former the displacement or force control and measurements achieved are not elaborated in sufficient detail.

For the soft tissue loading devices mentioned above the force measurement systems included the application of static weights [24], repetition of the experiment outside the MRI environment [22], the application of electric force sensors which suffer from electromagnetic interference [18] or force transducers outside the MRI room [23]. Other force sensors for the MRI environment have also been developed such as piezoelectric sensors. However these do not allow static force measurements and may induce image artefacts [12]. Since for the current study intrinsically MRI compatible devices are of interest optical force sensors are the focus. MRI compatible optical force sensors have been proposed for analysis of needle deflection and force feedback measurement during MRI based catheterisation have also been developed [15, 30] however these are applied to relatively low forces in the range 0-0.5 N. Tokuno *et al.* [31] developed a force

sensor based on optical micrometry and demonstrated accuracy within 1.6% for forces up to 6N. Tada *et al.* 2005 [32] presented a tri-axial force sensor based on optical micrometry consisting of 5 optical fibres (1 emitting and 4 receptor fibres) and demonstrated under 3% errors for forces up to 15 N. However when in a later study [26] the authors used the system for human finger-tip indentation the force, acquired at 1 Hz, was found to vary with a standard deviation of up to 0.3 N around a static mean load of 2.35 N. The causes for these increased deviations, with respect to the calibration, were not discussed. Recently Song *et al.* [14] designed an advanced Fibre Bragg Grating (FBG) based tri-axial force sensor mounted on a robot arm. Forces were calibrated up to ~10 N and a maximum force error of 0.5 N was recorded.

In this study a FBG sensor system is proposed because these have several advantages over other optical force sensing systems [14]: 1) *measurement is independent of fluctuating light levels*, 2) *multiple gratings can be applied in series* and 3) *they have simple wiring and a compact implementation*.

This paper outlines the design of a complete system for soft tissue indentation which is MRI compatible. This system is computer controlled and incorporates dynamic displacement control and motions can be triggered to be synchronised with the imaging (enabling the use of repeated image acquisitions common for MRI based deformation measurement techniques). Embedded in the MRI actuator is a novel high speed (100 Hz) MRI compatible optical FBG based force sensor enabling dynamic force measurement suitable for viscoelastic (e.g. ramp and hold type) analysis and fast detection of timing of motion and load application. The system is evaluated for the MRI based investigation of soft tissue biomechanics based on phantom tests and indentation of the upper arm of volunteers.

2. Methods

This section describes: 1) *Fibre Bragg Grating Based Optical Force Sensing*, 2) *The soft tissue indenter system*, 3) *Optical force sensor calibration* and 4) *Evaluation of the indenter system performance*. All signal and image processing methods were developed in MATLAB (R2012a v7.14.0.739, The Mathworks Inc., USA).

2.1. Fibre Bragg Grating Based Optical Force Sensing

In Fibre Bragg Grating (FBG) a periodic perturbation of the refractive index is introduced along an optical fibre acting as a local wavelength specific reflector [33, 34]. The reflected (Bragg) wavelength λ_B for a specific grating is defined by [35]:

$$\lambda_B = 2\eta_{eff}\Lambda \quad 2-1$$

Here η_{eff} is the effective refractive index of the fibre core and Λ is the period of the grating. From this equation it is clear that λ_B is both strain and temperature dependant since η_{eff} varies with temperature and Λ is altered following longitudinal fibre strain and thermal expansion/contraction [35]. Under isothermal conditions a linear relationship exists between reflected wavelength and the

applied strain where tensile and compressive strains increase and decrease the wavelength reflected respectively. For the current study only the mechanical strain induced effect is of interest since it is linearly dependent on the force exerted on the optical fibre. In order to separate the effect of mechanical strain from the effects of temperature fluctuations two gratings are placed close together in series whereby one is subjected to both mechanical strain and local temperature variations, while the other is isolated from mechanical strain and acts as a (temperature) reference grating. The wavelength reflected from the latter reference grating λ_T is thus purely a function of temperature and, together with the wavelength reflected from the former (strain) grating λ_S , can be used to derive the mechanical fibre strain ε as [36]:

$$\varepsilon = \frac{1}{S_\varepsilon} \left[\log \left(\frac{\lambda_S}{\lambda_{S0}} \right) - \log \left(\frac{\lambda_T}{\lambda_{T0}} \right) \right] \quad 2-2$$

Here $S_\varepsilon = 1.2 \pm 0.03 \text{ pm}/\mu\varepsilon$ represents the strain sensitivity. The subscript 0 denotes initial values which can be derived for an unloaded fibre prior to each testing session by taking the mean signal (here 1000 samples across 10 seconds is used) for both gratings.

For the current study a high strength optical fibre (GeO₂ doped silica glass fibre, ORCOMER® coated, 125 μm in diameter) containing two FBG gratings (Draw Tower Grating pair, FBGS International, Belgium) was used. The nominal Bragg wavelengths for the gratings used are 1532 nm and 1530 nm for the strain and reference grating respectively (hence as tensile strain is applied the reflected wavelength increases away from the reference grating wavelength). The reflected wavelength peaks were acquired at 1 kHz using an optical interrogator (SM130-700, Micron Optics Inc., USA). However for the current study the data was stored using a 10 point data interleave (running average of 10 consecutive data samples) resulting in an effective measurement frequency of 100Hz.

Due to the brittle nature of the fibre material (silica glass) it is best to load fibres in tension rather than compression. In tension the fibres used are capable of supporting loads up to 50N (corresponding to a 5% breakage strain). This is sufficient for the current study since measurement of forces up to 15 N are of interest (i.e. loads occurring during mild indentation of soft tissue).

2.2. The soft tissue indenter system

FIG. 1 shows an overview of the complete soft tissue indentation system. In the control room (left in FIG. 1) a computer (FIG. 1.A) linked with data acquisition units and custom electronics (FIG. 1.B) controls the motion of a hydraulic master cylinder (FIG. 1.C) whose motion is linked with the MRI compatible indenter assembly (FIG. 1.D) in the MRI room (right in FIG. 1). The force sensor data is acquired using an optical interrogator system (FIG. 1.E). This section discusses the soft tissue indenter system in the following steps: 1) *The MRI compatible soft tissue indenter assembly*, 2) *The MRI actuator assembly*, 3) *The indenter head and force sensor assembly*, 4) *Actuator motion control and data acquisition*. For each section the most relevant parts are discussed with reference to an associated figure.

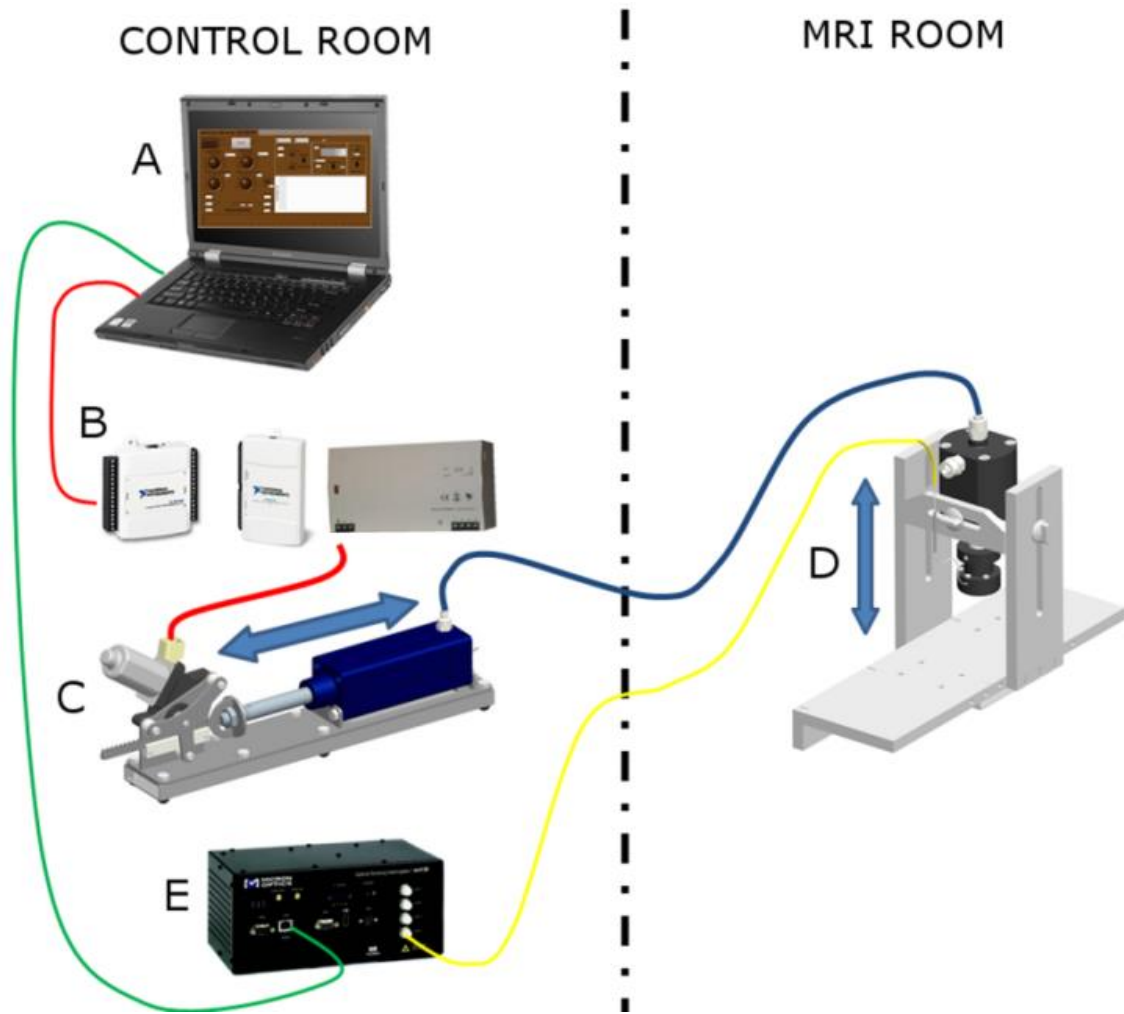


FIG. 1. Overview of the MRI compatible indenter and control system

2.2.1. The MRI compatible soft tissue indenter assembly

The design for the MRI compatible soft tissue indenter assembly will now be discussed with reference to FIG. 2. The MRI actuator body (1) is mounted on a support bridge (2) which is attached to two side plates (3) that are mounted onto the bottom plate (4). The whole assembly can be fixed on the scanner bed using a slide rail (5) and support ridge (6). The actuator orientation (blue arrows) can be adjusted using the adjustment screws (7) and (8) and slot (A) to set desired loading angle and maximum depth. The maximum indentation is set by placing the piston in its maximum deployed position, moving the actuator down until a desired maximum tissue indentation is reached and then securing the adjustment screws on the support bridge and side plates. The bridge set-up shown in FIG. 2 can be used for indentation of extremity soft tissue sites such as the biceps region of the upper arm or the tibialis anterior region of the lower leg. For other tissue regions a different mounting structure can easily be incorporated. All parts are made of polyamide except for the MRI compatible actuator body which is discussed in the following section.

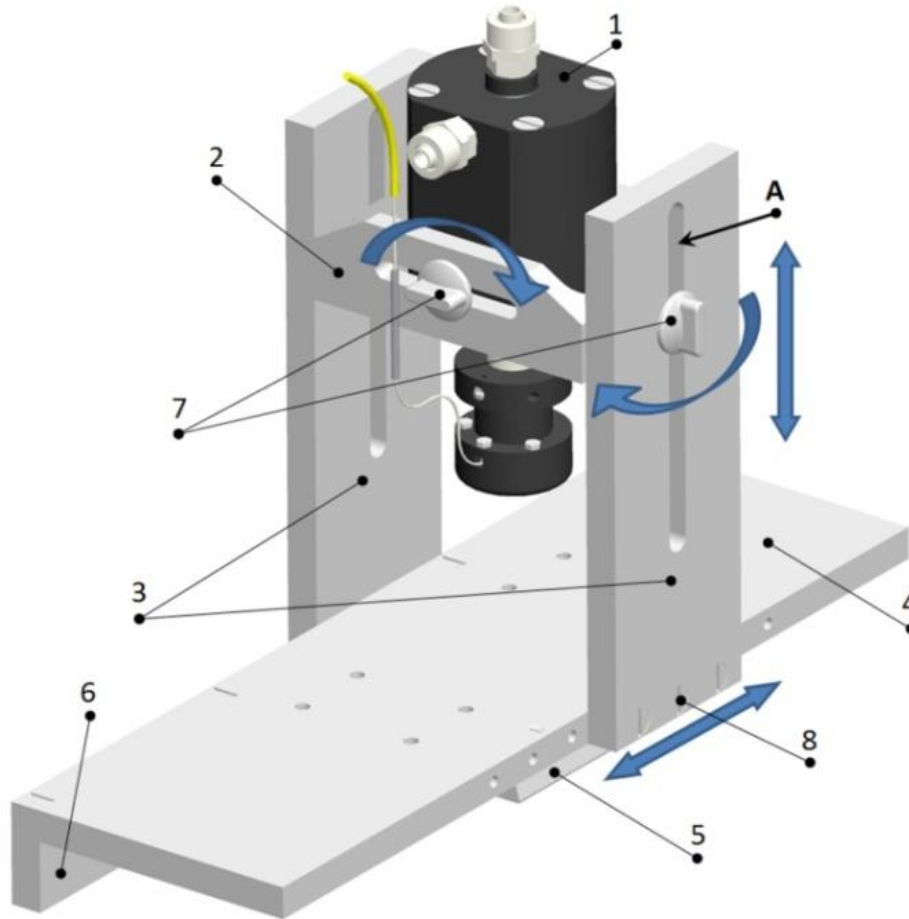


FIG. 2. The MRI compatible indenter assembly

2.2.2. The MRI actuator assembly

This section discusses the MRI actuator assembly with reference to FIG. 3. The tube connector (1) allows water driven by the master cylinder to enter the actuator chamber formed via parts (2) and (3). The tube connector (4) and the rubber seal (5) form a manual valve which can be opened to allow for air removal during filling of the system. Parts (2) and (3) clamp a soft rubber diaphragm (6) which deforms under the influence of water pressure as the piston shaft (7), attached via the ring (8) is pushed downwards (see right side of FIG. 3). The maximum stroke is 44 mm. The use of a diaphragm ensures low initial friction. The piston contains a flat face such that it can be constrained from rotating using part (9). The bottom part of the piston contains the indenter head and force sensor assembly (10) which is discussed in the following section. All black parts in FIG. 3 are constructed of polyoxymethylene and all white parts are polyamide except for the piston which is made of polytetrafluorethylene to ensure low friction.

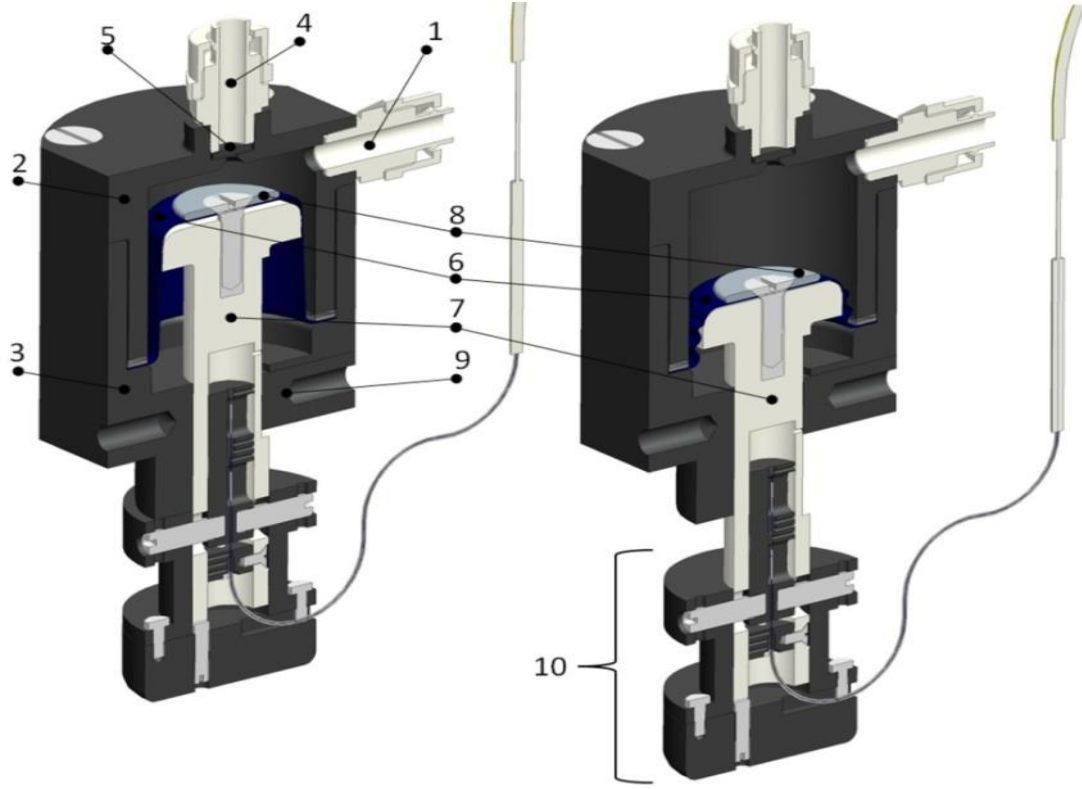


FIG. 3. The MRI actuator assembly in a retracted (left) and outward (right) state

2.2.3. The indenter head and force sensor assembly

The indenter head and force sensor assembly is discussed with reference to FIG. 4. The bottom of the piston shaft (1) is inserted into the piston head parts (2-3) which are able to slide relative to the piston shaft by an amount limited by the set screw (4). The piston head is flat and 45 mm in diameter with rounded edges 6 mm in radius. This head size was chosen since it provides a sufficiently large compression site relative to bulk soft tissue regions such as the biceps region of the upper arm. Using set screws (5) the piston head parts (2-3) are attached to part (6) of the force sensor assembly (right) which is inserted into the piston shaft (1). Part (7) of the force sensor assembly is fixed to the piston shaft via the screw (8). The FBG sensor fibre (9) enters the piston head at site (A) where it is supported using a bolt (10). The fibre reference and strain gratings (8 mm long each and 18 mm apart) are located at sites (B) and (C) respectively. The fibre (9) runs through a central hole in parts (6) and (7) where it is glued (EPO-TEK 353ND, Epoxy Technology Inc., USA) at sites (D) using the glue injection holes at (E). Thus when a compressive force F_C is applied to the bottom of the indenter head it slides with respect to the piston shaft, converting the load to a tensile force F_T at the strain grating (C) while the temperature reference grating (B) remains unloaded. The force F_C is directly proportional to the force F_T . Therefore due to the current design in the case of downward indentation with respect to gravity the weight of the indenter head components (0.94 N) first needs to be overcome. Hence for downward indentation only forces in excess of this weight can be recorded. Pre-tension can be

introduced in the fibre using a screw (11) which also ensures that the weight of the indenter head assembly does not buckle the fibre.

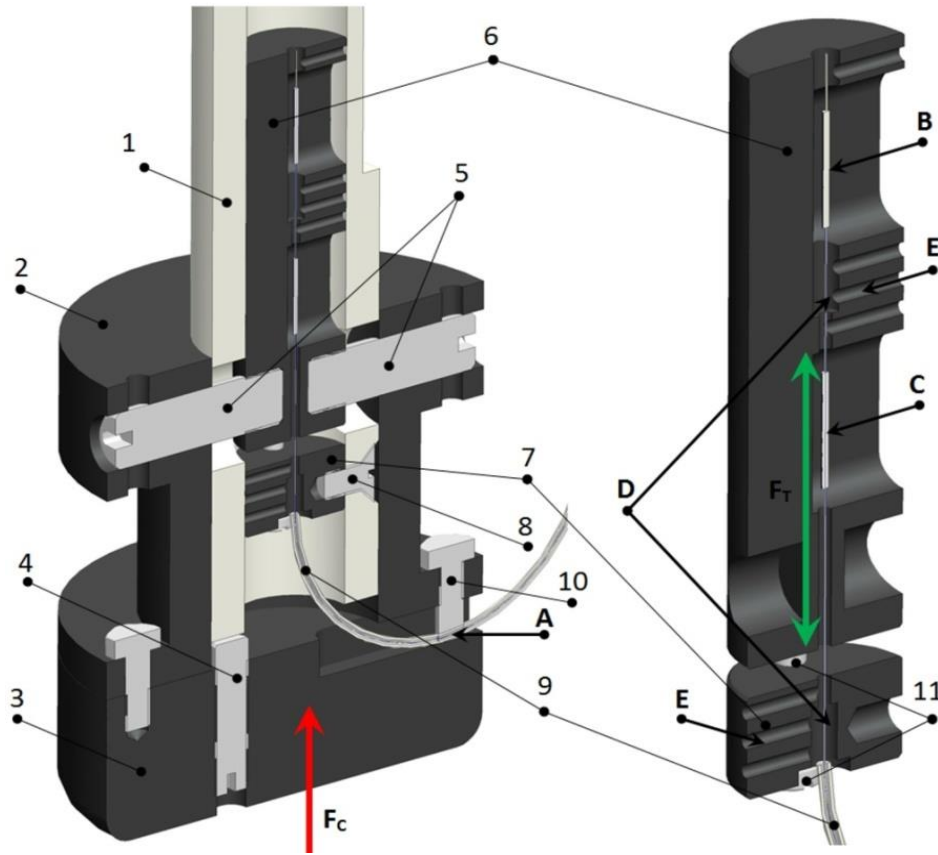


FIG. 4. The indenter head (left) and force sensor assembly

2.2.4. Actuator motion control and data acquisition

The motion of the MRI compatible actuator discussed above is dictated by a hydraulic master cylinder assembly (see also FIG. 1C) placed outside the MRI room, which will be discussed using FIG. 5. A 24 V DC-motor (1) (FBG 0130821708, Bosch, UK) powers the movement of a steel hydraulic cylinder (2) (CP95SDB50-80, SMC Corporation, USA) via an attached stainless steel gear rack (3) forcing water in or out through the valve (4) which is linked to the MRI compatible actuator via a 12 mm diameter polyurethane tube. The motion of the master cylinder (and thus the MRI actuator) is regulated through computer control (see also FIG. 1A-C) of the DC-motor. Custom control hardware and software (LabVIEW 8.6, National Instruments Corporation, USA) incorporating two data acquisition units (FIG. 1B) (USB 6009 and USB 6211, National Instruments Corporation, USA) were used.

The DC-motor has an in-built gear assembly with an internal to external gear-ratio of 62:1. The internal axis contains two 90° offset Hall sensors. Counting of the Hall sensor pulse edges (rising and falling for both sensors) thus provides 248 position references per external axis rotation which translates into approximately 1 positional reference for each 0.3 mm of MRI actuator motion. Prior to testing the positional reference is calibrated by first powering the DC motor to move the MRI actuator to its minimum (fully retracted) position and then to its final (maximum indentation depth)

position. This provides a “map” of the indenter position for its full range of motion. The DC motor power and speed are adjusted through pulse width modulation of the input voltage. During the repeatable indentations the pulse width is ramped up to a set maximum (to achieve desired maximum motor power and speed), then as the indenter approaches its final position the pulse width is ramped down to a set minimum (for instance to achieve a desired minimum motor power during a hold phase) when the indenter approaches its set final position. Therefore in the current study a pulse width profile is prescribed controlling motor power and position (force or speed control can also be implemented).

Since the indenter location, orientation and final depth are set by hand prior to scanning (see section 2.2.1) the actual applied indentation depth and displacement history need to be derived from the image data. FIG. 6 demonstrates the location, depth and orientation of the indentation can all be verified from the MRI data (segmentation based on face normal analysis of the triangulated skin iso-surface). The deformation history can then be derived by mapping the DC motor position history with respect to the segmented final displacement of the indentation.

The LabVIEW based control and acquisition software provides a graphical user interface enabling: 1) *setting of motor power parameters*, 2) *position control and speed monitoring*, 3) *setting of FBG parameters and acquisition of FBG data over time (based on Transmission Control Protocol communication with optical interrogator, see FIG. 1E) and 4) motion triggering toward an MRI scanner generated Transistor-Transistor-Logic (TTL) trigger pulse*. LabVIEW was run on a laptop PC, with a 32-bit Microsoft Windows Vista Business operational system, 3.5 Gb RAM, and dual core 2.1 GHz processors.

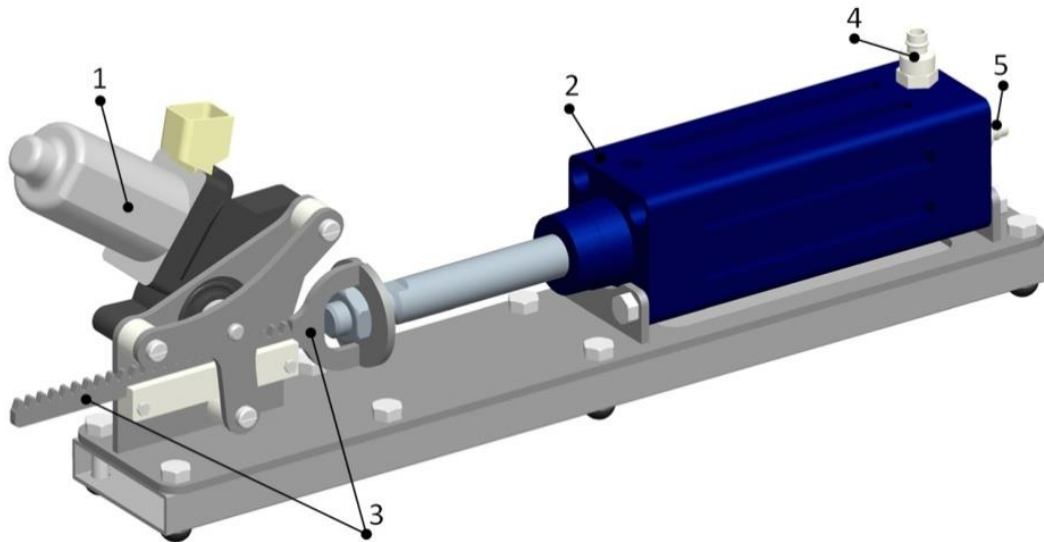


FIG. 5. The hydraulic master cylinder assembly

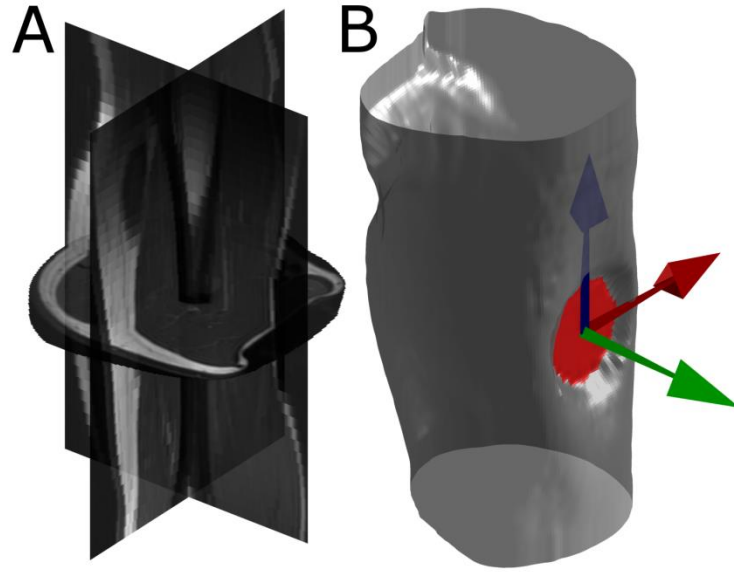


FIG. 6. Sample anatomical MRI data-set for an indented tissue site (upper arm) (A) from which the indentation site (red surface in B) can be segmented for verification of indentation location, depth and orientation (arrows in B show the indenter coordinate

2.3. Optical force sensor calibration

In order to relate the FBG derived fibre strain measure (equation 2-2) to fibre force, uniaxial compression testing was done using a Zwick Z005 (Zwick GmbH & Co., Germany) equipped with a 50 N load-cell. The system was subjected to three types of force controlled loading in the range of 0-15 N: 1) a stair case test (1 N steps at 0.25 N/s, followed by 30 s hold phase, repeated 3 times), 2) a stair case test (1 N steps at 0.75 N/s, followed by 30 s hold phase, repeated twice) and 2) high speed ramp loading (ramp up to 15 N at 5 N/s followed by 10 second hold and ramp down to 0 N). A 0.5 N pre-load was used for all tests.

The staircase tests were used for calibration by segmenting the plateaus regions for both the load-cell and FBG strain curves. In order to determine a mapping from the FBG strain to force the average plateau forces for each all tests (75 points, 15 for each of the 5 tests) were used simultaneously. Both linear (scaling based) and non-linear (cubic spline based) mappings were investigated. Following calibration based on the staircase tests, the force curve for the high speed ramp loading was then predicted using the stair-case based calibration and was thus effectively used for verification purposes.

2.4. Evaluation of the indenter system performance

The intended application for the indenter is the non-invasive investigation of soft tissue mechanical properties whereby during indentation all boundary conditions required for inverse analysis are recorded. These boundary conditions include: 1) the indentation force derived from the FBG sensor, 2) the complex soft tissue deformation acquired using SPAMM tagged MRI [2, 3], 3) the tissue geometry

for the construction of FEA models derivable from anatomical MRI and 4) muscle tissue fibre architecture obtained from diffusion tensor MRI [1] to allow for analysis of anisotropic material behaviour.

In order to evaluate indenter system performance for the above application it was applied (FIG. 7) for indentation of a silicone gel phantom [37] and the upper arm region of volunteers (ethical approval and informed consent obtained from the Medical Ethical Committee, Academic Medical Centre, Amsterdam, The Netherlands). All the above mentioned boundary conditions were recorded during the experiments. However only the measurements relevant to the indenter system performance are highlighted in detail here as the soft tissue deformation measurements were presented elsewhere [2, 3] and are beyond the scope of this paper. Since the quality of these measurements also illustrates the utility of the indenter system they are briefly summarized in the discussion section.

The indenter system was evaluated in the following ways: 1) *force measurement within the MRI environment*, 2) *Evaluation of indenter motion repeatability*, 2) and 3) *Evaluation of MRI compatibility*.

All scans were performed on a 3.0 T scanner (Philips Intera, Philips Healthcare, Best, The Netherlands) and FIG. 7A-B highlights coil placement of the flexible surface coils used (Flex-M, Philips Healthcare, Best, The Netherlands) and FIG. 7C-D shows iso-surface visualisations of the indented configurations and sample SPAMM tagged MRI slices. In order to compute 3D soft tissue deformation SPAMM tagged MRI data was acquired in three mutually orthogonal directions [2, 3]. However for both the phantom and volunteer data one of the three SPAMM tagged MRI directions was used for demonstration of indenter performance.

For the phantom tests repeated indentations were performed while a dynamic series ($n = 60$) of SPAMM tagged MRI data were acquired (1-1 SPAMM, 123 ms delay, 177 ms 3D Transient Field Echo read-out, T_R/T_E 2.53/1.28 ms, field of view 120x120x39 mm, acquisition matrix 80x52, 10 slices, reconstructed voxel size 0.93x0.93x1.5 mm). Timing of indenter motion with respect to imaging resulted in 11 image dynamics per indentation cycle and 5 complete repeated indentations per dynamic series.

For the volunteer tests repeated indentations were performed while a dynamic series ($n = 250$) of SPAMM tagged MRI data were acquired (1-1 SPAMM, 100 ms delay, 177 ms 3D Transient Field Echo read-out, T_R/T_E 2.42/1.19 ms, field of view 120x120x40 mm, acquisition matrix 80x60, 10 slices, reconstructed voxel size 0.94x0.94x2 mm). Timing of indenter motion with respect to imaging resulted in 71 image dynamics per indentation cycle and 3 complete repeated indentations per dynamic series.

Indentation cycles were triggered using a TTL pulse timed to start with the first dynamic or the first dynamic following completion of a deformation cycle.

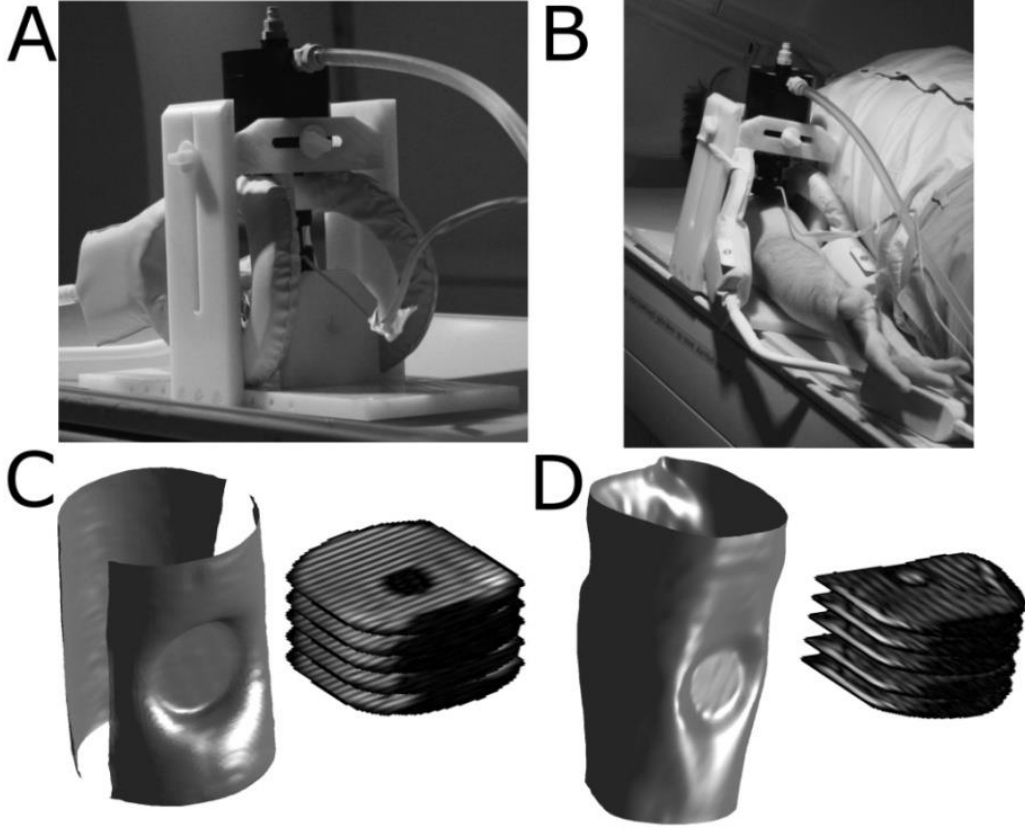


FIG. 7. MRI compatible indenter system positioned for indentation of the silicone gel phantom (A) and the upper arm of a volunteer (B). Coils are placed laterally to the regions of interest and can be secured to the side plates. In addition iso-surfaces illustrating the indentation sites and selected SPAMM tagged MRI data slices are shown for the phantom (C) and volunteer (D).

2.4.1. Force measurement within the MRI environment

During the indentation experiments the FBG derived force was recorded at 100 Hz. Since the force sensor is optical fibre based no interference with the MRI imaging is expected. Since skeletal muscle tissue is highly viscoelastic [38, 39] it is expected that, for the indentation tests of ramp and hold type discussed above, force relaxation should be observed during the hold phase. To demonstrate the sensor's performance in the MRI environment for application to soft tissue biomechanics, its ability to register this viscoelastic force history is demonstrated for two load rates of 10 mm/s and 20 mm/s.

2.4.2. Evaluation of indenter motion repeatability

In order to study repeatability of the indenter motion, the dynamic SPAMM tagged MRI series (see FIG. 7C-D) for the phantom ($n = 60$ dynamics) and for the volunteer ($n=250$ dynamics) were analysed. The SPAMM tagging encodes for motion and are thus well suited for repeatability analysis. Each dynamic series can be represented as a 4D data set $N_{i,j,k,d}$ where (i, j, k) represent voxel (row, column and slice) indices and d the index for dynamics. In case a repeated periodic motion occurred during the dynamic series some of these dynamics are thus equivalent to each other with the exception of differences induced by noise. In order to establish whether the indenter motion was indeed

repeatable a $n \times n$ temporal sum of squared differences (SSD) matrix (SSDM) was created for both the phantom and volunteer data where all dynamics are compared to all others using:

$$SSDM_{p,q} = \sum_{i,j,k} (N_{i,j,k,p} - N_{i,j,k,q})^2 \quad 2-3$$

where each entry in the matrix at (p, q) reflects the sum of squared differences of dynamic p with respect to dynamic q for a random selection of 10000 voxels within the gel/tissue volume. The matrix SSDM is obviously symmetric around its diagonal where all entries are zero since here $p = q$. However if a repeated periodic motion occurred during the dynamic series other parallel diagonals with minimal differences are to be found. For instance if a repeated motion with period x (dynamics) occurred then multiple diagonal minima exist since the following entries in SSDM should all reflect only differences due to noise: $[(p, p + x), (p + 1, p + 1 + x), (p + 2, p + 2 + x), \dots]$. For the current study a period of 11 and 71 (see description of indentation experiment at the start of section 2.4) should therefore be observable in the data for the phantom and volunteer data respectively (e.g. for the phantom data dynamic 1 is repeated at dynamic 12 and 23, etc.). Analysis of parallel diagonal locations in the SSDM, showing difference magnitudes expected for noise (i.e. similar to differences between multiple static repetitions) thus allowed for demonstration of repeatability.

2.4.3. Evaluation of the system MRI compatibility

The MRI environment poses significant design challenges for the safe and appropriate functioning of both the device and the MRI scanner. In this study the following definition of MRI compatibility is used (for current definitions of MRI safety terminology see [7, 40]. Although the term “MRI compatibility” is no longer favoured by the ASTM it is commonly used in the literature and hence also adopted here): *A device or system is Magnetic Resonance (MR) compatible if, when used in the MR environment, is MR safe and has been demonstrated to neither significantly affect the quality of the [MRI data or its] diagnostic information, nor have its operations affected by the MR device.* Since all of the device features presented, that are to be used in the MR environment are non-conducting (with the exception of the tap water used in the hydraulic system), non-metallic and non-magnetic the indenter system operation is not significantly affected by the MRI scanner and can be termed MR safe using scientific rationale [7]. In addition all materials employed in the MRI environment (e.g. polyoxymethylene, polyamide, polytetrafluorethylene and polyurethane) exhibit appropriate magnetic susceptibility [41] for MRI and thus it is likely that their influence on MRI data quality is minimal. Nonetheless to evaluate the effect of the indenter on MRI data quality, system performance was analysed in the MRI environment. MRI data was acquired for a silicone gel phantom (FIG. 7). In order to study the effect of the indenter presence on image quality a dynamic series ($n=100$) of MRI data was acquired with and without the indenter present. The same sequence as above was employed however without the SPAMM pattern (177 ms 3D Transient Field Echo read-out, T_R/T_E 2.38/1.15 ms, field of view 120x120x40 mm, acquisition matrix 80x60, 20 slices, reconstructed voxel size 0.94x0.94x2 mm). To study the effect of the presence of the indenter for each dynamic series,

per voxel temporal signal to noise (SNR) ratios $tSNR$ were derived for each data set (with and without the indenter) using:

$$tSNR_{ijk} = \frac{\overline{\mu_{ijk}}}{\overline{\sigma_{ijk}}} \sqrt{n} \quad 2-4$$

where $\overline{\mu_{ijk}}$ and $\overline{\sigma_{ijk}}$ represents the mean and standard deviation of voxel (i, j, k) respectively across all dynamics.

3. Results

3.1. Optical force sensor performance

FIG. 8 shows the staircase test force curves used for calibration (3 repetitions for 0.25N/s and 2 for 0.75N/s) and overlain the results for fitting based on the FBG strain signal. The current configuration of the sensor can only measure forces in excess of indenter head assembly weight. Hence all FBG derived force curves presented here start at 0.94N. Using linear scaling (scale factor 970.20) followed by shifting (adding 0.94 N to compensate for the weight of the indenter head assembly) the FBG strain could be linearly related to force (FIG. 8A). Although this showed an overall good correlation (FIG. 8B) with the load-cell force ($R^2=0.99$), force differences up to 0.16 N (corresponding to 3.2% difference with respect to 5 N) occurred and a maximum difference percentage of 5.6% was encountered (corresponding to a 0.056 N difference from 1 N). However all curves (FIG. 8A) demonstrate a small degree of non-linearity of the FBG force sensor system. Linear mapping thus led to underestimation for low forces where the curve was initially slightly concave followed by overestimation of larger >7N forces due to a mild convex curvature (see curvature in FIG. 8B). Hence to take the mild non-linearity into account a cubic spline fit (MATLAB function *CSAPE*) was used to map the FBG strain to force (FIG. 8C) producing a piecewise-polynomial form relating FBG-strain to force. The fit was constrained at the ends such that the end slopes match the slope of the cubic of the last 4 data points to allow for reasonable extrapolation in forces in the range 15-50 N (up to breakage). The cubic spline based mapping of the FBG strain to force, reduced the maximum error percentage to 3.1% (corresponding to a 0.031 N difference from 1 N) and the maximum force error magnitude to 0.043 N (corresponding to 0.7% difference with respect to 6 N). The mean and standard deviation of the differences with respect to the load-cell force were 0 N ($-2.23 \cdot 10^{-5}$ N) and 0.015 N respectively. FIG. 8B shows the high degree of correlation (FIG. 8D) between the load-cell force and the FBG derived force ($R^2=1.00$). The largest percentage difference coincided with the (1 N) first plateau measurements which is closest to 0.94, the lowest recordable force. If for all tests the first plateau is ignored, the maximum error percentage is 1.2% (corresponding to a 0.024 N difference from 2 N).

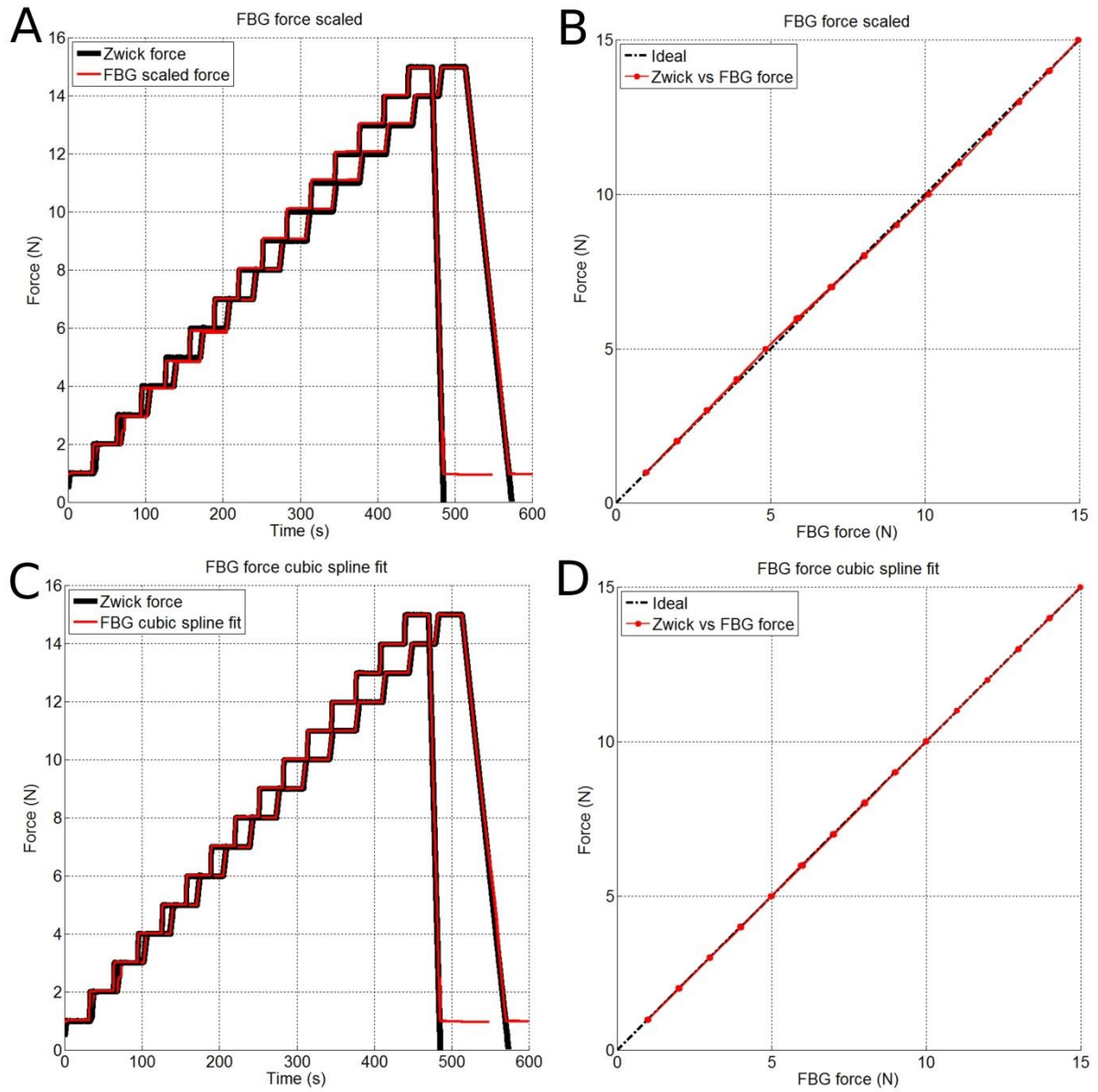


FIG. 8. The load-cell (thick black) and, overlain, the FBG derived (red) force history curves for the stair-case tests (A-C) and, the load-cell and FBG derived plateau forces plotted against each other to show correlation (B-D). Images (A-B) are for linear scaling while (C-D) are for cubic spline based mapping of FBG-strain to force.

After calibration based on the stair-case tests the response for a ramp test at 5 N/s was predicted (FIG. 9). The differences with respect to the load-cell force were not found to increase for the higher load rate which is also apparent from the large degree of overlap in FIG. 9.

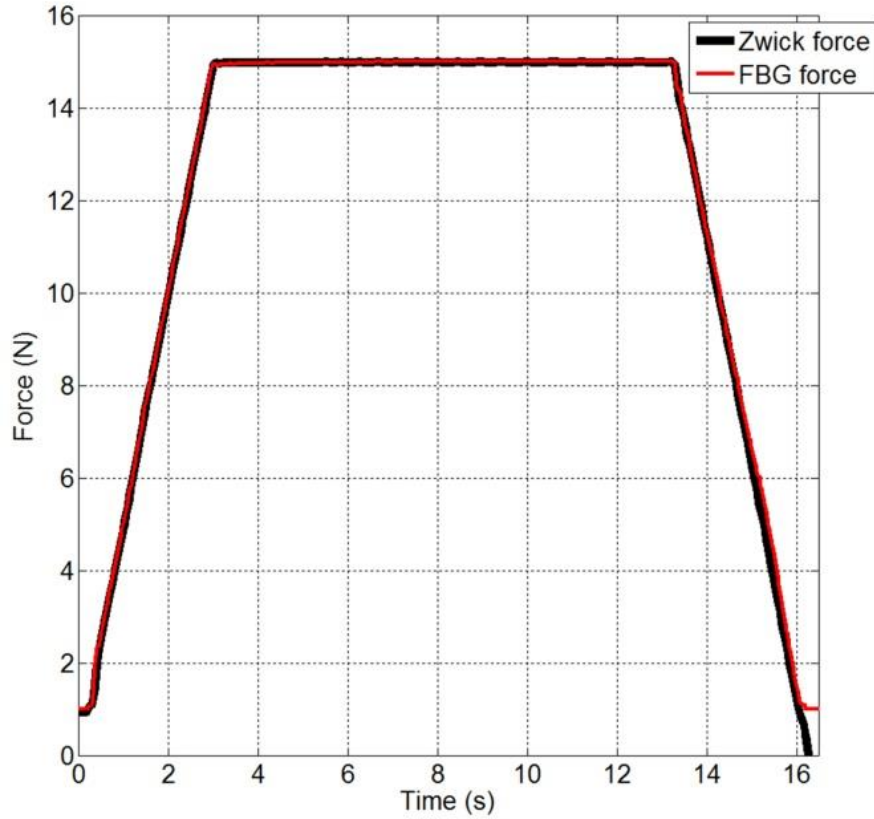


FIG. 9. The load-cell (thick black) and, overlain, the FBG derived (red) force history

FIG. 10 shows sample force and displacement history curves for volunteer upper arm indentation within the MRI scanner. As expected, no signal interference from the MRI scanner was observed. Both sets of curves in FIG. 10 demonstrate viscoelastic force decay during the hold phase of the load. In addition preconditioning is observed (FIG. 10B) as the peak force is reduced during the first motion cycles. This is not a repeatability artefact but a well-known empirical finding for biological soft tissue. The loading rate for the first volunteer in FIG. 10A (a male volunteer aged 30) was twice as high for the second volunteer in FIG. 10B (a female volunteer aged 25). Besides indentation depth and subject to subject variation it is likely that the increase in viscoelastic response observed in the second volunteer can be explained by the increase in loading rate.

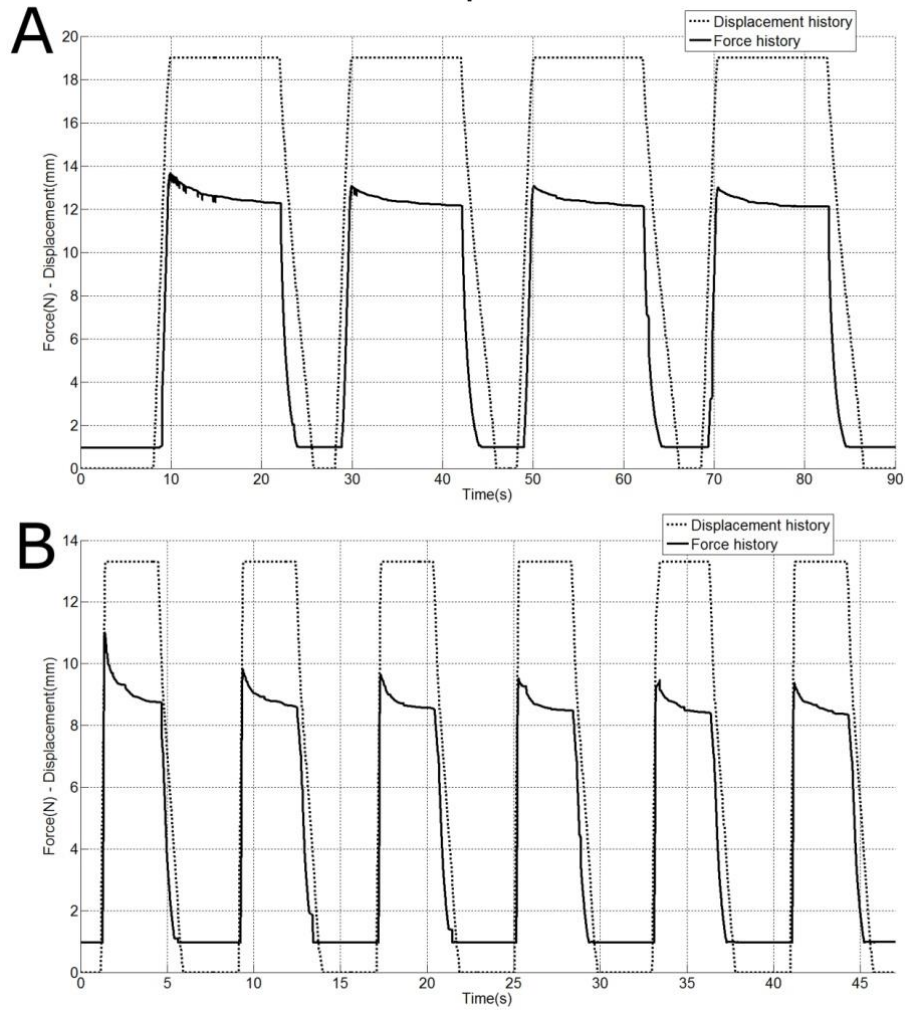


FIG. 10. The sample displacement (dashed) and force (solid) curves for cyclic ramp and hold indentation tests for two different volunteer tests. Part (A) shows 19mm indentation at 10mm/s and a 12 s hold phase. Part (A) shows 13.3 mm indentation at 20mm/s and a 3s hold phase.

3.2. Repeatability of indenter motion

FIG. 11 shows a visualisation of the sum of squared differences matrix (SSDM) for the phantom and volunteer data (due to symmetry only half of the matrix is shown). As mentioned in section 2.4.2, periodic motions should be reflected in the SSDM as parallel diagonals showing minimal differences. These are visible for both the phantom and volunteer data. Each row in the SSDM describes a periodic variation starting on the left, on the main diagonal, with a zero magnitude difference (difference with respect to itself) the magnitudes alter but return to a minimum every 11th and 71st entry in the rows for the phantom and volunteer respectively. In order to visualise this periodic nature for all rows more appropriately FIG. 11B and E show shifted or synchronised SSDMs. Here, starting at the bottom, each row of entries in the SSDM was shifted to the left by an amount equivalent to its corresponding row (or dynamic) number minus 1 (e.g. first row remains the same, second is shifted by 1, third by 2 etc.). This shifting effectively aligns or synchronises all the periodic difference variations in the

SSDMs. The numbered labels in FIG. 11B and E now clearly show that for the phantom and volunteer data the periodic signal exactly reflected the desired periods of 11 and 71 dynamics respectively. This is further illustrated in FIG. 11C and F which are obtained by taking the mean of images FIG. 11B and E in the row direction. As FIG. 11B and E show clear minima exist at the appropriate locations (red dots) and differences rapidly increase to more than an order of magnitude higher for non-repeated dissimilar images. It was found that for static repetitions during the hold phases the mean and standard deviation of the SSD were $1.13 \cdot 10^6$ and $0.25 \cdot 10^6$ for the phantom data and $4.14 \cdot 10^4$ and $0.60 \cdot 10^4$ for the human data. These thus form the SSD values expected purely due to noise variations. The SSD values for acquisitions during indenter motion, which, according to applied indenter motion, should be equivalent, showed a mean and standard deviation of $1.14 \cdot 10^6$ and $0.52 \cdot 10^6$ for the phantom data and $4.37 \cdot 10^4$ and $0.94 \cdot 10^4$ for the human data. Hence it may be concluded that the indenter presents with a highly repeatable motion since the differences expected due to noise are of equivalent magnitude to the differences between repeated indenter motions.

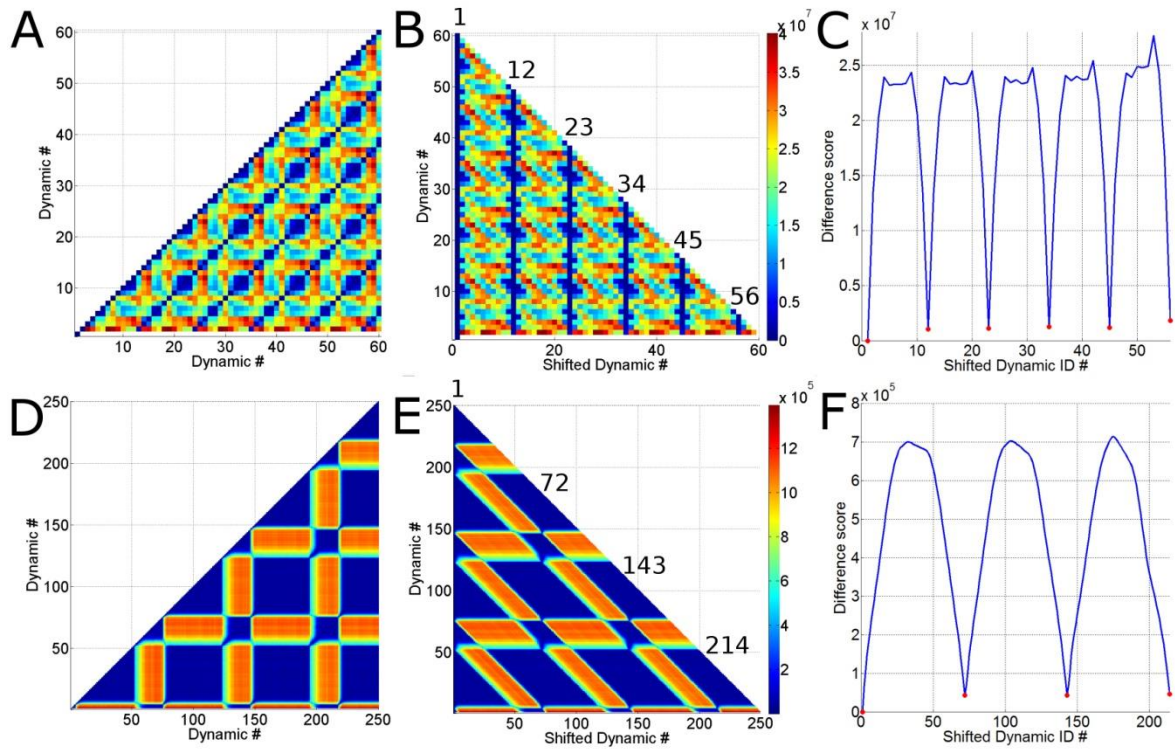


FIG. 11. The motion cycle repeatability results for the phantom (A-C) and volunteer (D-F). (A-D) show the SSDM's. (B-E) show the shifted or "synchronised" SSDMs and (C-F) show derived mean SSD graphs (blue line) were red dots indicate the cycle repetition points.

3.3. MRI compatibility of indenter system

FIG. 12 below shows temporal SNR images derived for the phantom without (A) and with indenter present (B). It was found that throughout the field of view the temporal SNR levels were not

significantly altered by the presence of the indenter. The overall mean and standard deviation of the temporal SNR levels for the phantom volume without indenter present were 500.18 and 207.08 respectively. When the indenter was added this became 501.95 and 200.45 respectively. This result was invariant under permutation of read-out directions and or slice orientations. However since the MRI actuator contains tap water, which presents with signal, certain read-out directions may present with fold-over due to this signal. However this can easily be avoided by alternating the read-out direction or by employing fold-over suppression techniques.

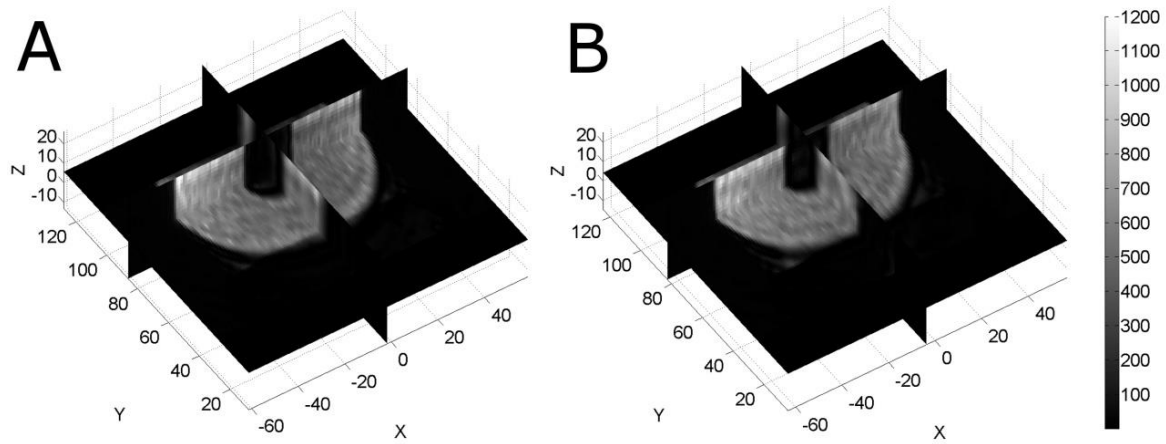


FIG. 12. The temporal SNR images for the phantom without (A) and with indenter present (B).

4. Discussion

A novel MRI compatible soft tissue indenter system and optical FBG based force sensor have been presented. The computer controlled indenter motion is highly repeatable since MRI acquisitions during repeated indenter motions did not induce significant additional variation on top of what is expected due to noise. In addition the indenter device and force sensor are fully MRI compatible and are manufactured from non-ferromagnetic materials. The MRI compatibility was also evident following evaluation inside an MRI scanner and no negative effects such as SNR decrease and or image artefacts were observed.

Many current soft tissue loading devices for the MRI environment do not have dynamic force displacement control capabilities and have been limited to quasi-static deformation analysis [18, 21, 22, 24, 26, 28, 29], the evaluation of isotropic hyperelastic constitutive models [18, 21, 22, 24, 26, 28, 29], estimation of strain from 2D imaging [21, 27, 28] or from finite element simulations [21, 22]. Force measurement is sometimes based on the application of static weights [24] or by repeating the experiment outside the MRI environment [22] or electric force sensors which suffer from MRI scanning induced electromagnetic interference [18]. Recently Fu et al. [27] used an MRI compatible indenter system to study leg tissue biomechanics and 2D strain estimates were derived for 7 frames per indentation cycle based on harmonic phase MRI. However the actuator control and force sensing capabilities were not discussed in detail, hindering comparison to the current study. Solis et al. 2012 developed an advanced soft tissue loading system capable of motion control and triggering to MRI acquisitions. However the system (designed for porcine buttock compression) cannot be applied to

upper arm indentation and is currently limited to loading parallel to the MRI bore axis. To the authors' knowledge this is therefore the first study to demonstrate a computer controlled MRI compatible indenter with an integrated high sampling rate (100 Hz) force sensor suitable for the non-invasive analysis of the complex anisotropic, viscoelastic and 3D mechanical behaviour of human skeletal muscle tissue.

The current study employs Fibre Bragg Grating for MRI compatible force measurement and calibration demonstrated force difference percentages no larger than 3.1% for forces up to 15 N and the maximum overall force difference magnitude was 0.043 N. The maximum force difference percentage occurred close to the minimum force measurement of 0.94 N. If forces in excess of 1 N are considered the maximum force difference percentage is reduced to 1.2%. MRI compatible optical force sensors in the literature for the range of 0-15 N present with errors in the order of 1.6% ([31] 0-6 N) up to 3% ([32] 0-15 N, 1 Hz) and Song *et al.* [14] presents FBG based force sensing with force errors up to 0.5 N for loads up to 10 N. The force sensor performance demonstrated in the current study is therefore comparable (sometimes better) to that reported in the literature. Evaluation of the force sensor during soft tissue indentation demonstrated its ability to record viscoelastic force histories resulting from ramp and hold indentations, see FIG. 10.

As referred to in section 2.4 the indenter system was evaluated for use in the non-invasive investigation of tissue biomechanics. Indentation tests were performed on a silicone gel phantom and the upper arm of volunteers. During indentation all boundary conditions required for inverse FEA analysis of the biomechanical tissue properties were acquired such as: 1) *the indentation force derived from the FBG sensor*, 2) *the complex soft tissue deformation acquired using SPAMM (SPAtial Modulation of the Magnetisation) tagged MRI [2, 3]*, 3) *the tissue geometry for the construction of FEA models derivable from anatomical MRI and 4) muscle tissue fibre architecture obtained from diffusion tensor MRI [42] to allow for analysis of anisotropic material behaviour*. FIG. 13 shows the application of the indenter system to indentation of a volunteer upper arm. Following application of SPAMM tagged MRI in three mutually orthogonal directions dynamic 3D deformation could be measured (FIG. 13A-B). In addition diffusion tensor MRI allowed for characterisation of fibre architecture throughout the region of interest (FIG. 13C) and anatomical MRI scans provided the basis for the construction of detailed FEA models (FIG. 13D). Recently the MRI compatible indenter set-up was used for validation of SPAMM tagged MRI based 3D soft tissue deformation measurement in a silicone gel phantom and the upper arm of a volunteer [2]. The repeatable indenter motion allowed for the quantification of deformation measurement accuracy and precision at sub-voxel levels, further demonstrating the high level of repeatability of the indenter motion.

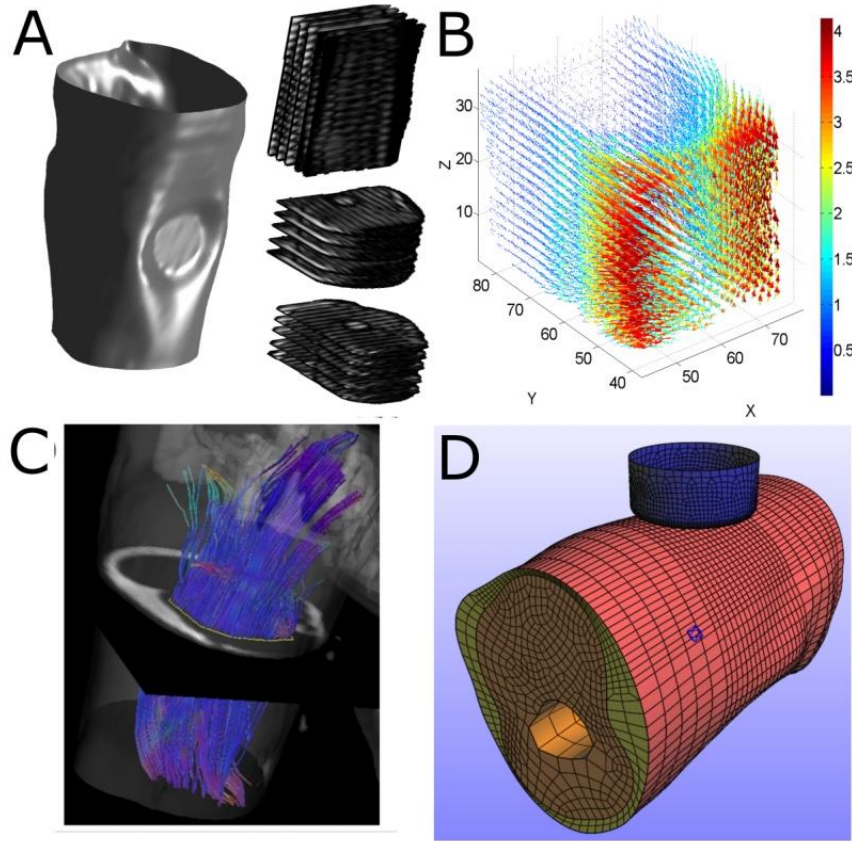


FIG. 13. SPAMM tagged MRI acquired during indentation (A) allowing derivation of 3D dynamic deformation measurements (units mm) (B). Diffusion tensor measurements provide fibre architecture (C) which together with anatomical MRI data allows for the construction of detailed finite element models (D).

FIG. 14 shows two alternative applications for the MRI compatible indenter system, large strain mechanical property investigation based on the combined use of SPAMM tagged MRI and MR Elastography (see FIG. 14A and the preliminary study [43]) and for validation of motion compensation techniques for dynamic contrast enhanced imaging (FIG. 14B). The indenter has also found application outside the MRI environment for inverse mechanical property analysis combined with digital image correlation [44].

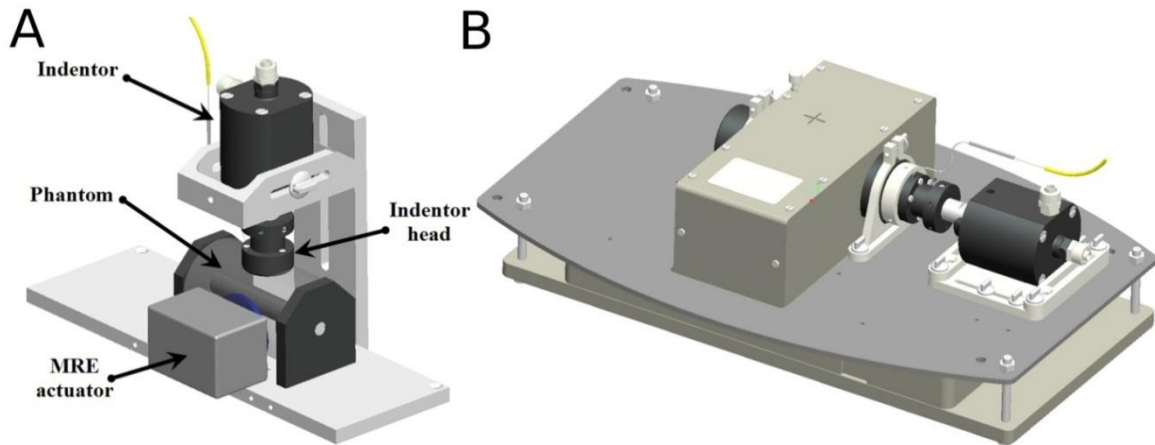


FIG. 14. Example alternative applications of the MRI compatible actuator. For large strain MR Elastography (MRE) measurements (A) and mounted for use with a specialised coil and phantom for validation of motion compensation techniques during contrast enhanced imaging (B).

Some limitations of the proposed MRI compatible indenter and force sensor system are now discussed. The indenter system employs a hydraulic (tap water driven) master slave system. If imaging is performed using read-out directions aligned with the water filled actuator signal fold-over may occur. However this can easily be dealt with using alteration of the read-out direction, field of view adjustments or fold-over suppression techniques.

At present the indenter stroke is limited to 44 mm. This is sufficient for application to (non-injury inducing) soft tissue indentation in human subjects. For other applications requiring a larger range of motion the design can be scaled up.

Due to the way that compressive forces are converted to tensile forces in FBG sensor within the current design, measurement of forces in the range 0~0.94 N are not possible at present since the weight of the indenter head (0.94 N) assembly needs to be overcome (in the case of downward indentation). The force measurement range 0.94~15 N is however sufficient for many purposes including large strain biomechanical soft tissue investigation and indentation of the upper arm as presented here.

Other researchers have developed more complex tri-axial force sensor systems (e.g. [14, 31, 32, 45]) for MRI. However the uniaxial force measurements presented here ensured a simple and compact indenter design and are sufficient for comparison to inverse FEA as the same resultant uniaxial force can easily be generated as an output.

For the current study the Fibre Bragg Grating signal was acquired at 1 kHz however a 10 point data interleave was used for the optical interrogator leading to an effective acquisition rate of 100 Hz. Although 100 Hz is deemed sufficient for the applications of the current study, a sample rate up to 1 kHz is feasible with the employed optical interrogator (or higher using more advanced interrogators). However acquiring the sensor data at 1kHz was not possible in the current study given the limitations in computer speed and of the indenter control and data acquisition software used. The computer control system must simultaneously record the FBG signals and record and control the DC-motor behaviour and monitor the MRI TTL pulse in real-time. As such in the future force

measurement of up to 1 kHz will be possible with this indenter given improvements in computational power and improvements in the data acquisition software.

With the exception of the optical interrogator system the proposed MRI compatible indenter system is relatively low cost. The MRI compatible indenter slave system components (used within MRI environment) are fabricated from non-conducting non-ferromagnetic materials which are all common engineering polymers (e.g. polyoxymethylene). All non-standard components can be manufactured using simple mill and rotating bench operations and all screws, tubes, fittings, the DC-motor and master cylinder are commercially available. If readers are interested in the design specifications or computer aided design files (based on Creo 1.0, Parametric Technology Company, MA, USA) these can be made freely available upon request.

Together with MRI modalities such as SPAMM tagged MRI, the indenter system allows for analysis of all boundary conditions required for the non-invasive investigation of the complex mechanical properties of soft tissue. Future work will focus on the use of the proposed system for the analysis of the non-linear elastic, anisotropic and viscoelastic mechanical properties of skeletal muscle tissue with application to the field of impact biomechanics and pressure ulcer prevention. This is the first complete system for MRI based analysis of upper arm indentation in humans.

5. Conclusion

A novel MRI compatible indenter system is presented for the investigation of soft tissue biomechanics. A master slave system was developed whereby a computer controlled hydraulic master cylinder was used to provide highly repeatable motions to an MRI compatible actuator. To evaluate the system in the MRI environment and demonstrate its usefulness for soft tissue biomechanics investigation the system was used for indentation of a silicone gel phantom and the upper arm of volunteers. Repeatability was evident from the fact that MRI data for static repetitions showed similar variations as those from dynamic motion repetitions. All indenter components in the MRI room are non-ferromagnetic and non-conducting and hence MRI safe and can be used in close proximity to the imaging subject. MRI compatibility was demonstrated following imaging of a phantom and the presence of the MRI actuator did not induce any artefacts or significant SNR changes. Embedded in the indenter assembly is a novel high sampling speed (currently 100Hz) optical FBG based force sensor. The force sensor was calibrated for forces up to 15N and demonstrated a maximum force difference of 3.1% (maximum force difference magnitude was 0.043 N). Application of the force sensor in volunteer upper arm indentation showed the sensor's ability to register viscoelastic force decay resulting from ramp and hold indentation.

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Conflict of interest statement

The authors have no conflicts of interest to disclose.

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