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Three-Dimensional Ultrasound Strain Imaging of Skeletal Muscles

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***Abstract*—Muscle contraction is characterized by large deformation and translation, which requires a multi-dimensional imaging modality to reveal its behavior. Previous work on ultrasound strain imaging of the muscle contraction was limited to 2D and bi-plane techniques. In this study, a three-dimensional (3D) ultrasound strain imaging technique was tested against 2D strain imaging and used for quantifying deformation of skeletal muscles. A phantom compression study was conducted for an experimental validation of both 2D and 3D methods. The phantom was compressed 3% vertically and pre- and post-compression full volume radio frequency (RF) ultrasound data were acquired using a matrix array transducer. A cross-correlation-based algorithm using either 2D or 3D kernels was applied to obtain the displacement estimates. These estimates were converted to Cartesian space and subsequently, strain was derived using a least-squares strain estimator (LSQSE). The 3D results were compared with the 2D results and the theoretically predicted displacement and strain. Comparison between 2D and 3D kernels was performed on data from a plane with a large tilt angle to study the influence of out-of-plane motion on the two techniques. To demonstrate the *in vivo* feasibility, 3D strain was calculated from live 3D data, acquired during a 2 second isometric contraction and relaxation of the quadriceps muscle in a healthy volunteer. The phantom study showed good correlation between estimated displacements and the theoretically predicted displacements. Root-mean-squared errors (RMSE) were 0.16, 0.17 and 0.13 mm in the x-, y- and z-direction respectively. The absolute RMSE for the 3D strain values were 0.94, 1.2 and 0.41% in the x-, y- and z-direction respectively. The 2D method performed worse, with 3 (x-direction) to 6 (z-direction) times higher RMSE values. The larger errors in lateral and elevational direction with respect to the axial RMSE are potentially caused by the large angle between the ultrasound beams. Initial *in vivo* results revealed 3D strain curves which clearly visualized the contraction and relaxation of the quadriceps muscles. Muscle deformation estimation using real-time 3D ultrasound RF-data seems feasible and the use of 3D kernels improves displacement estimation in comparison to 2D techniques. Future work will focus on improving lateral and elevational displacement estimation, and investigating local differences of strain in skeletal muscles and its clinical relevance.**

Keywords—3D; strain, skeletal muscle; out-of-plane motion;

I. Introduction

The aetiology of many musculoskeletal diseases is related to the biomechanical condition of the affected muscle. Techniques which can provide systematic and quantitative data of abnormal tissue deformation could be of great diagnostic value for treatment of musculoskeletal diseases or injuries. Muscle contraction is characterized by large translations and deformations, which requires a multi-dimensional imaging modality to reveal its deformation. Strain imaging or elastography is a widely used technique for measuring passive and active deformation of tissue.

Lopata et al. performed a bi-planar acquisition to assess deformation of the biceps during contraction in three orthogonal directions [1]. Although this imaging technique is fast and yields data of two orthogonal planes, errors in the strain estimation might be introduced by out-of-plane motion. A full 3D imaging technique could reduce these errors and quantify the complex behavior of muscles due to their anisotropic material properties and contractile element orientations. Full 3D strain imaging has already been used in cardiac applications [2] but the improvement of 3D-based over 2D-based strain estimation has not been investigated yet.

The goal of this study is to demonstrate the use of 3D ultrasound for strain estimation in skeletal muscles, and investigate its potential advantages over 2D-based strain estimation. 2D and 3D US acquisitions were performed in a deformable phantom for experimental validation. The 3D technique was applied to measure muscle deformation *in vivo* of the quadriceps muscles on a healthy volunteer.

II. Materials & Methods

A. 3D Ultrasound Acquisition

Three-dimensional radio-frequency (RF-data) ultrasound data were acquired with an iE33 ultrasound system (Philip Medical Systems, Bothell, USA), equipped with an X7-2 matrix array transducer (2-7 MHz) and an RF-interface which sampled the RF-data with a frequency of 16 MHz.

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B. Phantom Study

A phantom experiment was conducted for validation and comparison of 2D vs. 3D displacement and strain estimation. A phantom block (10x10x10 cm^3) was constructed from a homogeneous 10% gelatin (Dr. Oetker, Ede – The Netherlands) solution. 1% Silica scattering particles (15-30 µm SiC, E. Merck, Darmstadt – Germany) were added to mimic the scattering properties of soft tissue [3]. The block was assumed to be linearly elastic and nearly incompressible (Poisson's ratio 0.495). Using an automated plate compressor, the phantom was compressed vertically 3 mm, resulting in 3% strain in z-direction and 1.5% strain in the x- and y-direction. The transducer was attached to the compressor plate and captured 3D 'full volume' data before and after the compressive deformation. The scanning angle between the image lines was 1° in both lateral and elevation direction.

As a reference (i.e. ground truth), the displacements and strains within the phantom were derived using classic continuum mechanics; Using Hooke's law the displacement and strain at every point were calculated given the applied compression. Absolute root-mean-squared errors (RMSE) for both techniques were calculated from the estimations and the theoretical ground truth. Comparison between 2D and 3D kernels was performed on data from a central plane (elevation angle: $\Phi = 0°$) and from a plane with a large tilt angle (elevation angle: $\Phi = 32°$) to study the influence of out-of-plane motion on the two techniques (Fig. 1).

C. In Vivo: Quadriceps Muscle

To demonstrate the feasibility of 3D strain estimation of skeletal muscles *in vivo*, 3D ultrasound RF data were recorded during a voluntary isometric contraction of the quadriceps muscle group (upper leg) of a healthy volunteer. The subject was instructed to contract and relax the muscle during a two second cycle. During the *in vivo* studies the imaging mode was set to 'Live 3D' resulting in 1° and 1.25° angles between the lines in lateral and elevational direction respectively. The imaging depth was set to 6 cm, resulting in a volume rate of 39 Hz. The transducer was positioned at approximately 2/3 distally of the upper leg, with the lateral axis of the transducer aligned parallel to the femur bone (Fig. 2).

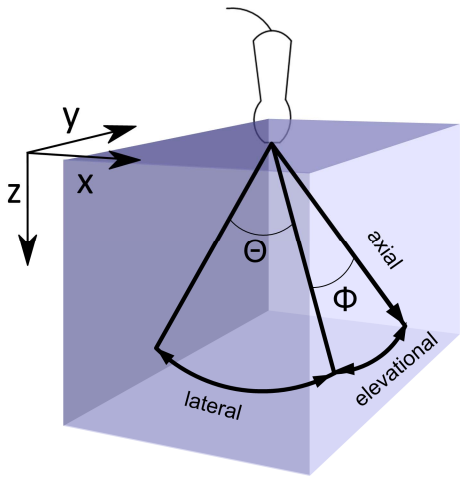

Fig. 1. Phantom block and representation of both the transducer and Cartesian coordinate system.

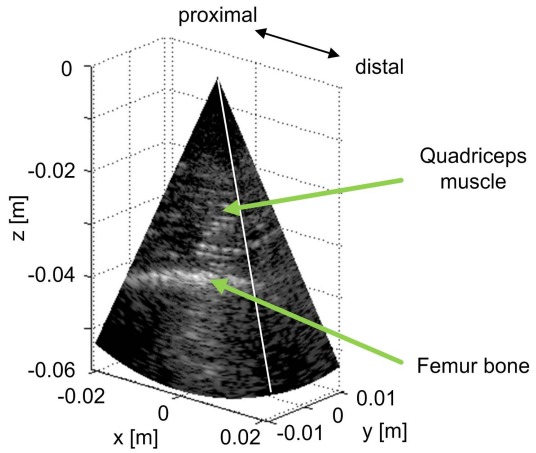

Fig. 2. Three-dimensional ultrasound image of the quadriceps muscle group.

D. Displacement and Strain Estimation

Two multi-level cross-correlation-based methods were compared for their performance of displacement and strain estimation. The first method cross correlated 2D segments of ultrasound RF-data [4]. The second method used 3D segments for the cross correlation [2]. The multi-level approach allows a high resolution and local displacement estimation by decreasing the size of the data segments to the desired resolution in multiple iterative steps. For a fast and accurate sub-sample displacement estimation the cross correlation peak was detected using parabolic fitting.

Inter-frame displacement estimates were filtered with a median filter to remove outliers. The displacement estimates accumulated over all frames were converted to Cartesian space and subsequently strain was derived using a least-squares strain estimator (LSQSE) [5]. Settings for the displacement estimation are summarized in Table 1.

TABLE I. USED SETTINGS FOR EVERY LEVEL OF THE DISPLACEMENT ESTIMATION ALGORITHM

	Iteration 1	***Iteration 2***	***Iteration 3***
Used data	Envelope	Envelope	Envelope
Axial segment size	5 mm	2.5 mm	1.2 mm
Axial search range	7.2 mm	0.6 mm	0.15mm
Axial shift (overlap)	1.25 mm (75%)	0.62 mm (75%)	0.28 mm (75%)
Lateral segment size	9 lines	5 lines	5 lines
Lateral search range	5 lines	3 lines	3 lines
Elevational segment size*	9 lines	5 lines	5 lines
Elevational search range*	5 lines	3 lines	3 lines
Axial median filter	5.0 mm x 5 lines		
Lat/Ele* median filter	5.0 mm x 11 lines		

* Only applicable for the 3D method

III. Results

A. *Phantom Study*

Figure 3 shows the measured displacements and strain values of the homogeneous phantom for an applied strain of 3% using the 3D method. Overall the measured displacements are in accordance with the theoretically predicted values. Assuming incompressibility of the phantom (i.e. Poisson's ratio = 0.495) the strain in the x- and y-direction must be two times smaller than the strain in the z-direction, which is corroborated by the results of the strain estimates.

RMSE values of the displacement and strain estimates were calculated to assess the performance for both the 2D and 3D method. Without out-of-plane motion ($\Phi = 0°$), the RMSE values of displacement estimation for the 3D method were 0.13, 0.07 and 0.07 mm in the x-, y- and z-direction respectively. Since there is no out-of-plane motion the 2D displacement estimation show similar results with RMSE values of 0.21 and 0.17 mm in the x- and z-direction. The RMSE values for the 3D strain estimates were 0.93, 1.2 and 0.41 % in the x-, y- and z-direction respectively. The RMSE values for the 2D strain estimates were 1.2 and 0.58 % in the x- and z-direction.

At a large elevation angle ($\Phi = 32°$) there is out-of-plane motion due to the expansion of the phantom in the x- and y-direction. For this situation, the RMSE values of displacement estimation for the 3D method were 0.16, 0.17 and 0.13 mm in x-, y- and z-direction respectively. The RMSE values for the 3D strain estimates were 0.94, 1.2 and 0.60 % in the x-, y- and z-direction respectively. The RMSE values of the 2D method, affected by the out-of-plane motion, were 3 to 6 times higher in x- and z-direction respectively.

B. *In Vivo: Quadriceps Muscle Mroup*

The 3D method was used to derive strain within the quadriceps muscle group. Figure 4a depicts the strain curves in x-, y- and z-direction during a voluntary isometric contraction and relaxation cycle of 2 seconds. The images showing strain in the z-direction for three annotated time points in Figure 4a are depicted in Figure 4b. The strain curves show the global behavior of the muscle group. Strain in the z-direction reaches a maximum of 20%. At the end of the cycle, i.e. at relaxation of the muscle, we see a negative strain value in z-direction. This indicates that the ultrasound acquisition was not synchronous with the start of the contraction cycle.

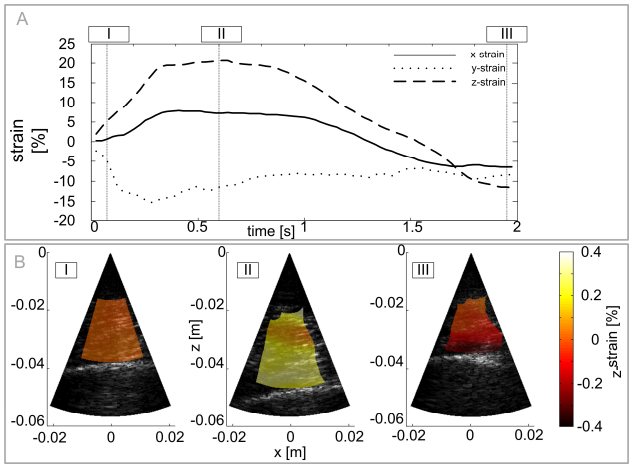

Fig. 4. A) Mean strain curves as a function of time , and B) The strain in the z-direction at the middle b-mode image for the three annotated time points in panel A: point I, at starting of the contraction, point II, at maximum strain value and point III, at total relaxation of the muscles.

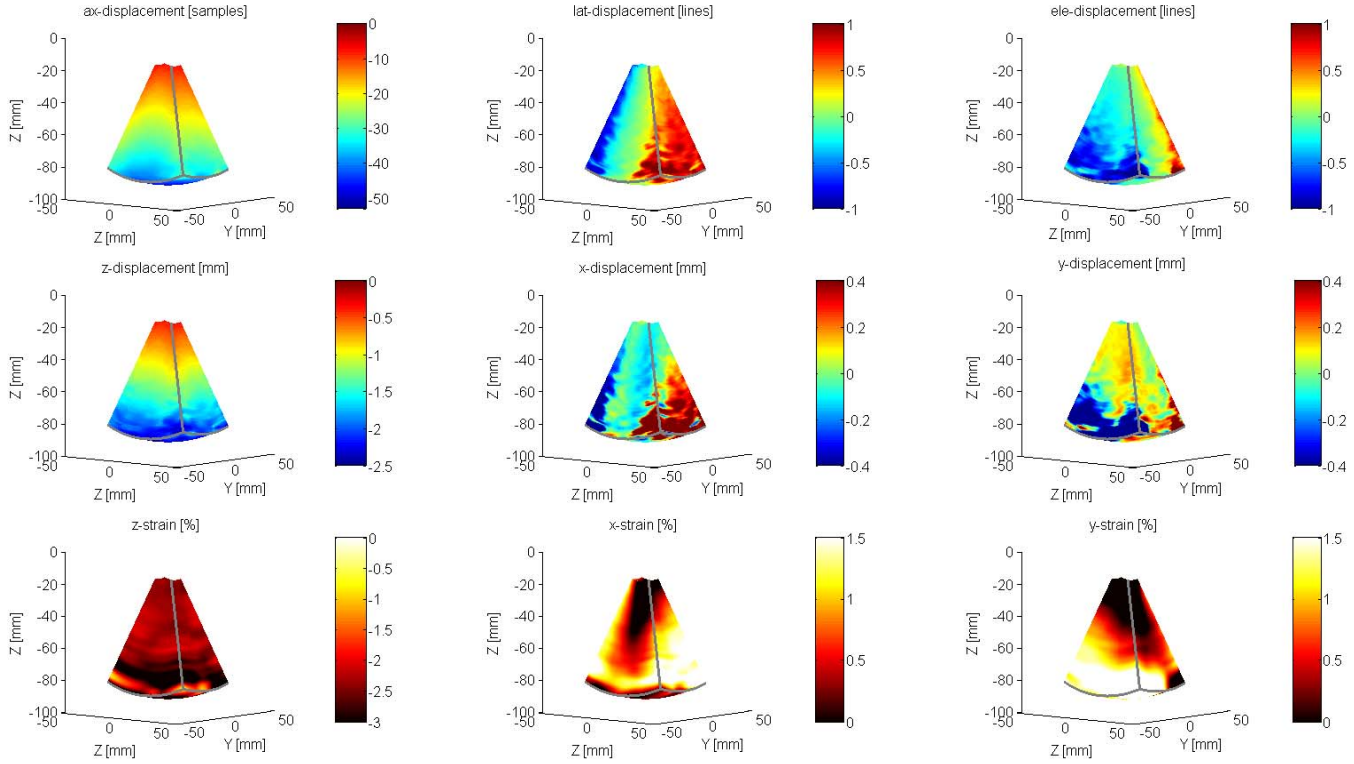

Fig. 3. Three-dimensional displacement and strain estimations of a 3% vertically compressed phantom. The top panel shows the displacement in axial, lateral and elevational direction. The middle panel shows displacement after conversion to x-, y- and z-direction. The lower panel shows the results of strain estimates using a 3D LSQSE.

IV. Discussion and Conclusion

Full volume 3D ultrasound imaging enables the quantitative measurement of complex tissue motion. Despite a lower spatial resolution and frame rate, 3D imaging has shown to be a valuable tool to quantify tissue motion, especially when there is motion in three directions, which results in out-of-plane motion for 2D techniques.

An important observation of this study is the good agreement of displacement and strain estimation in z-direction with the theoretically predicted values, but a relatively high RMSE for the x- and y-direction. This is probably caused by the low line density of the used transducer. Luo et al. [6] showed that a smaller pitch, a wider beam width and spline interpolation are required in order to reduce the error of the lateral displacement estimation. Additionally, the conversion from displacements in the transducer coordinate system to Cartesian coordinates reduces performance of the displacement estimation even further [7].

Overall, the results indicate that the use of 3D full volume data to estimate tissue displacement outperforms the use of 2D data. Please note that for the results, obtained by the 2D displacement method, 2D data (single planes) were selected from the 3D datasets acquired with the matrix array transducer. This transducer has lower spatial resolution (i.e. lower sampling frequency and line density) than dedicated transducers for 2D acquisition. It is therefore interesting for future studies to quantitatively compare the performance of displacement estimation using a matrix array transducer and a linear array transducer.

The use of phantoms to validate displacement algorithms can be discussed, as the behavior of phantoms cannot be controlled perfectly. Hence, errors in the displacement estimation could be a result small heterogeneities in the phantom or incorrect experimental conduct. Furthermore, the physical properties of a phantom are of great importance; the speed of sound and the stiffness of the phantom must be controlled to ensure correct boundary conditions for the theoretical calculations. To this extent, in this study much care was taken to construct the tissue mimicking phantoms. In general, the resulting ultrasound images and displacement estimations suggest that the phantom experiments were conducted properly without any notable artefacts. To avoid the use of phantoms, one could use ultrasound simulations [8]. Using simulations, it is possible to control the experimental environment more accurate, but mimicking true soft tissue properties and physical behavior of ultrasound transducers is limited. Additionally, simulations introduce even more computational time, especially the simulation of 3D full volume data.

The preliminary results of the *in vivo* study of the quadriceps muscle group show that the 3D algorithm can measure deformations in human muscles in the leg. The derived strains in the z-direction, perpendicular to the contraction direction of the muscle, are increasing during the voluntary contraction. This observation indicates that the derived strain in z-direction is at least qualitatively sound, since muscles bulk during contraction. However, the strain curve in the x-direction can be discussed, as the curve indicates the remarkable muscle behavior of stretching along the contraction direction (x-direction) during muscle activation. This could be caused by complex muscle interactions of the different muscles in the quadriceps, but it is more likely that the strain estimation in this direction is severely affected by erroneous displacement estimation in the lateral and elevational direction. For further quantitative verification of the measured strains during muscle contraction EMG and force measurements [1], and cross validation using magnetic resonance techniques will be included in future studies. Possible differences of muscle behavior within healthy subjects and musculoskeletal patients may provide valuable information that can help diagnose and treat patients with affected muscles.

Concluding, this study showed that displacement estimation using 3D segments improves the axial displacement estimation with respect to using 2D segments. Application of the 3D technique *in vivo* is feasible and results in high quality strain images in the z-direction. However, future work is required to optimize the lateral and elevational displacement estimations with additional verification of *in vivo* measurements.

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