Dynamic imaging of skeletal muscle contraction in three orthogonal directions

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1Clinical Physics Laboratory, Department of Pediatrics, Radboud University Nijmegen Medical Centre, Nijmegen, The Netherlands; 2Donders Institute for Brain, Cognition and Behaviour, Centre for Neuroscience, Department of Neurology, Radboud University Nijmegen Medical Centre, Nijmegen, The Netherlands; 3Cardiovascular Mechanics Group, Department of BioMedical Engineering, Eindhoven University of Technology, Eindhoven, The Netherlands; and 4Research Institute MOVE, Faculty of Human Movement Sciences, VU University, Amsterdam, The Netherlands

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Lopata RG, van Dijk JP, Pillen S, Nillesen MM, Maas H, Thijssen JM, Stegeman DF, de Korte CL. Dynamic imaging of skeletal muscle contraction in three orthogonal directions. J Appl Physiol 109: 906–915, 2010. First published July 8, 2010; doi:10.1152/japplphysiol.00092.2010.—In this study, a multidimensional strain estimation method using biplane ultrasound is presented to assess local relative deformation (i.e., local strain) in three orthogonal directions in skeletal muscles during induced and voluntary contractions. The method was tested in the musculus biceps brachii of five healthy subjects for three different types of muscle contraction: 1) excitation of the muscle with a single electrical pulse via the musculocutaneous nerve, resulting in a so-called “twitch” contraction; 2) a train of five pulses at 10 Hz and 20 Hz, respectively, to obtain a submaximum tetanic contraction; and 3) voluntary contractions at 30, 60, and 100% of maximum contraction force. Results show that biplane ultrasound strain imaging is feasible. The method yielded adequate performance using the radio frequency data in tracking the tissue motion and enabled the measurement of local deformation in both the vertical direction (orthogonal to the arm) and in the horizontal directions (parallel and perpendicular to direction of the arm) in two orthogonal cross sections of the muscle. The twitch experiments appeared to be reproducible in all three directions, and high strains in vertical (25 to 30%) and horizontal (~20% to ~10%) directions were measured. Visual inspection of both the ultrasound data, as well as the strain data, revealed a relaxation that was significantly slower than the force decay. The pulse train experiments nicely illustrated the performance of our technique: 1) similar patterns of force and strain waveforms were found; and 2) each stimulation frequency yielded a different strain pattern, e.g., peak vertical strain was 40% during 10-Hz stimulation and 60% during 20-Hz stimulation. The voluntary contraction patterns were found to be both practically feasible and reproducible, which will enable muscles and more natural contraction patterns to be examined without the need of electrical stimulation.

FOR A LONG TIME, STUDYING dynamic muscle properties in humans was the exclusive domain of mechanical measurements (external force and movement) and electrophysiology (electromyography, EMG). Imaging of skeletal muscles has been applied for more than three decades using different modalities, including MRI and ultrasonography (23, 28, 29). Apart from imaging of muscle anatomy, also quantification of translational movement and local deformation of muscle tissue will be of interest in fundamental and clinical questions. For these purposes, the high temporal resolution of more recent ultrasound equipment enables the investigation of dynamic image sequences, which provides information on the changes that occur within the muscle during contraction (local strain) (20, 29). Hence, ultrasound imaging of muscles may give more insight in the process of force production and its dependency on muscle structure. This might be of use when studying force generation in primary muscle disorders such as muscular dystrophies, but also in connective-tissue diseases in which changes of the mechanical properties of muscle and connective tissue could influence force transmission.

Muscle contraction is known for its anisotropic behavior, i.e., the deformation of the muscle is different in different directions. The complexity of skeletal muscle strain has been examined and shown with two-dimensional (2D) and three-dimensional (3D) finite element modeling (2, 5, 24, 32). However, most reports on (ultrasound) strain imaging in skeletal muscles are limited to one or two dimensions and are, therefore, unable to reveal the 3D anisotropic behavior of the muscle. A multidimensional imaging modality is needed for a comprehensive mapping of muscle contraction. Since the distribution of muscle strain is known to be nonuniform (2), such imaging should yield estimates of tissue strain locally.

Ultrasound “elastography” was introduced by Ophir et al. (26) as a technique to estimate the mechanical properties of biological tissues and organs from the relative deformation, i.e., strain. Local displacement is assessed by correlating segments of ultrasound data acquired sequentially (25, 26). By calculating the first-order spatial derivative of the displacement field(s), “strain images” are obtained. Initially, the technique was used to visualize the distribution of strain within tissue under external compression to estimate the mechanical properties of tissues and organs (4, 12, 25), i.e., a passive deformation. More recently, strain imaging was also applied to data acquired in actively deforming tissue, such as the heart, to assess its function (6, 11). A limited number of studies have reported on strain imaging in skeletal muscles. One of the first in vivo efforts was performed with Doppler-based techniques (9). Speckle tracking methods using the raw, radio frequency (RF) data were applied to skeletal muscles, both ex vivo (33) and in vivo (16, 34). Several studies can be found on the ultrasonic measurement of the aponeurosis strain (21, 22). A recent study by Deffieux et al. (7) reported on 1D strain imaging, i.e., strain in the direction of the ultrasound beam, in 2D and 3D images of electrically stimulated muscles using an ultrafast (frame rate > 1,000 Hz) ultrasound technique. Also in the field of MRI, both tracking of displacement, as well as elastography of skeletal muscle have been reported (8, 35). A
study by Pappas et al. (27) revealed the nonuniform shortening of the musculus biceps brachii using cine phase contrast MRI.

Ultrasound has several advantages over MRI, such as the bedside applicability, cost effectiveness, and the high spatial and temporal resolution. However, there is a trade-off between spatial resolution, field-of-view, and temporal resolution, which is especially evident in 3D applications. RF-based techniques have an optimal performance for strains between 0.1% and 5.0%. Consequently, the frame rate has to be high to keep the frame-to-frame tissue deformation within that range. If the strain rate becomes too high, the accuracy and precision of the strain estimates will decrease rapidly due to significant decorrelation of the ultrasound data (3). Because the motion of the muscle and the deformation within skeletal muscles will be large, a relatively high frame rate of the equipment is required for ultrasound strain imaging. “Biplane” imaging is a fast semi-3D imaging technique (frame rate = 25–120 Hz) that is commercially available. Besides that a relatively high frame rate can be achieved, the major advantage is the ability of acquiring data from two orthogonal planes simultaneously, yielding both a cross-sectional and longitudinal view of the muscle. Hence, the deformation can be assessed in three orthogonal directions during muscle contraction.

The primary goal of this study was to develop a multidimensional, noninvasive technique to track rapid muscle motion and to measure local tissue deformation (i.e., strain) accurately in three orthogonal directions. A biplane strain imaging method is proposed that uses the RF echo-data to estimate tissue deformation. The method is applied to the human m. biceps brachii during both voluntary and electrically induced contraction. The latter will result in a controlled reproducible contraction, whereas the former will be easier to perform and be less demanding in clinical practice.

MATERIALS AND METHODS

Experimental setup. Five healthy male subjects, free of any neurological complaints, were recruited (age 22–44 yr) and gave their written informed consent to participate in the study. The study conformed to the Declaration of Helsinki and to the local ethics committee.

Each subject was asked to lie down on a bed, with the left arm flexed in a 45° angle as shown in Fig. 1. The wrist was cuffed and attached to a force sensor. The force sensor and the wrist clamp were fixed onto a table. To minimize movements, the entire arm was supported by a vacuum pillow, which is a large synthetic bag containing small Styrofoam balls that becomes rigid after removing the air content with a vacuum pump. To record surface EMG of the m. biceps brachii, two electrodes were placed in a so-called belly-tendon montage; the active electrode was placed at about the middle of the muscle belly, the reference electrode was placed on the distal tendon (10). The ultrasound transducer was positioned on top of the m. biceps brachii, orthogonal to the muscle surface (see Fig. 1). A water balloon was placed between the transducer and the skin to provide an offset for the transducer and to prevent transducer motion during muscle contraction. This offset was necessary to include a larger area of the m. biceps brachii in the imaged sector.

To be able to record reproducible muscle contractions, the musculocutaneous nerve innervating the m. biceps brachii and m. brachialis was stimulated electrically in part of the experiments. A single, square, electric current pulse of 100 µs was given in the axilla using a hand-held electrode. It elicited a so-called twitch in the muscle. The current strength was increased until the compound muscle action potential, recorded by the EMG electrodes, no longer increased in amplitude, indicating complete recruitment of all muscle fibers in the muscle.

Biplane ultrasound images and force exerted at the forearm were measured for three different types of muscle contraction:

1) Excitation with a single electrical pulse on the musculocutaneous nerve (as described in Experimental setup). This twitch contraction was repeated three times for reproducibility analysis.

2) Excitation of the elbow flexors with a train of five such pulses at a rate of 10 Hz and 20 Hz.

3) Voluntary contractions at 30, 60, and 100% of maximum voluntary contraction (MVC). First, the MVC force was determined for each person by asking the subject to fully contract the muscle twice for 1 s with 1 min rest in between. If the two contractions were similar, the maximum was taken as the MVC. After a short period of rest, the subjects were asked to follow a block-shaped curve of 2 s that was shown on a monitor (Fig. 2). The force exerted by the subject was drawn on top of the predefined pattern for visual feedback. The subjects were familiarized with the procedure until they were capable of following the block-shaped curves on screen at 30%, 60%, and 100% of their MVC. Each subject practiced the predefined pattern at 30% MVC at least three times prior to the recording.

As stated above, results from voluntary contractions are less reproducible than stimulated contractions, but they might become clinically more relevant since they can be performed in virtually any skeletal muscle.

![Fig. 1. Schematic representation of the experimental set-up. This set-up enabled simultaneous excitation of the muscle, measurement of the exerted force, and EMG, as well as ultrasound data acquisition of the m. biceps brachii. The m. brachialis is not visible in this figure. The upper arm was fixed in a vacuum pillow, and the echo transducer was placed in a clamp. The force sensor and wrist brace were fixed to the table. The musculocutaneous nerve was stimulated electrically, resulting in muscle contraction.](Image)
The m. biceps brachii was chosen because it mainly flexes the forearm during stimulation of the musculocutaneous nerve, which produces a force that can easily be measured. We looked also at the tibialis anterior muscle as an alternative, but this muscle produces both dorsiflexion and inversion of the ankle, making it difficult to measure exerted force. Furthermore, we choose the m. biceps brachii for its superficial location and good accessibility for ultrasound imaging.

**Data acquisition.** Surface EMG data were digitized at 2,000 Hz (Refa 64, 22 bits, 71.5 nV/bit; TMSi, Oldenzaal, The Netherlands). Force data were recorded using a force transducer (AST, KAP-E 2KN, 2.5 mV/N; A.S.T. Gruppe, Wolnzach, Germany), digitized (2,000 Hz) using a data acquisition board (National Instruments, DAQpad 6015, 16 bits, 0.3 mV/bit; National Instruments, Austin, TX), and stored on the hard disk of a PC. An externally generated synchronization signal was recorded simultaneously with the EMG and force data and was fed into the ECG input of the ultrasound machine for off-line synchronization.

The ultrasound data were acquired using a real-time 3D ultrasound system (Philips iE33; Philips Medical Systems, Bothell, WA) equipped with an X7 matrix array transducer and an RF interface. The matrix array transducer, with a bandwidth of 2–7 MHz, was used in biplane mode at a frame rate of 38–50 Hz, depending on the size of the image sector. The RF data were sampled at 32 MHz and transported to an external hard disk using a USB 2.0 connection. With biplane imaging over a separate longitudinal or cross-sectional view, two orthogonal images in both the cross-sectional and longitudinal direction of the muscle were measured simultaneously (Fig. 3). This has the advantage that such semi-3D data can be used to evaluate tissue deformation in three orthogonal directions. The position of the ultrasound probe was carefully chosen to minimize out-of-plane motion in both views. All data were checked during postprocessing for large translational movement by comparing the first and final frame of each data set. The 2D tracking algorithm corrected for the remaining translational movement within the image plane.

**Biplane strain imaging.** The RF data were processed using a 2D displacement estimation and tracking algorithm that was tested on tissue mimicking phantoms and applied successfully in vivo on cardiac data (14, 18, 19). This algorithm uses cross-correlation of 2D segments of RF data to measure the displacement in two directions simultaneously. Within a 2D ultrasound sector scan image, the direction parallel to the ultrasound beam is called the axial direction, whereas the direction orthogonal to the ultrasound beam is known as the lateral direction (Fig. 3).

The algorithm comprises a so-called coarse-to-fine approach (31). The algorithm starts at coarse scale using the demodulated RF data (“signal envelope”) to find the global motion of the tissue. A large segment of echo data of frame $f$ (7.2 mm × 11 lines), is cross-
correlated with a larger search area within the next frame \( f + 1 \) (17.0 mm \( \times \) 21 lines). The position of the peak of the cross-correlation is detected, and 2D parabolic interpolation of this peak is performed to obtain subsample displacement estimates. The resulting 2D displacement fields are up-sampled to higher resolution and used in the next iteration (Fig. 4).

In the next two iterations, smaller precompression windows of RF data are used (2.4 mm \( \times \) 5 lines and 1.2 mm \( \times \) 5 lines, respectively) with an overlap of 80%. By stepwise decreasing the axial window size and using the RF data, both resolution and precision of the displacement estimates are enhanced (18). The use of smaller data windows also enables accurate RF-based displacement estimation in high-strain areas (3). Hence, the used window size is reduced significantly for the next iteration. The postcompression window sizes were also decreased (7.3 mm \( \times \) 11 lines and 3.7 mm \( \times \) 11 lines), thereby limiting the search area. It must be noted that the lateral window size and resolution cannot be expressed in units such as millimeters, since it is depth dependent because of diverging distance between image lines. Strain values were obtained every 0.3 mm in the axial direction and for each RF line (i.e., lateral direction), corresponding to a horizontal resolution ranging from 0.15 mm for superficial structures to 1.0 mm for structures at maximum depth. The axial and lateral displacement data were filtered with a median filter of 3.0 mm \( \times \) 5 lines and 3.0 mm \( \times \) 11 lines, respectively.

After the third and final iteration, the axial and lateral displacements are used to track the tissue, yielding the coordinates of each pixel for all consecutive frames (15). The resulting coordinates are transformed to Cartesian coordinates (see Fig. 4). Two-dimensional, least-squares strain estimators with a window size of 11 \( \times \) 5 and 3 \( \times \) 15 pixels are used to obtain the strains (\( \varepsilon \)) in the vertical \( (z) \) and horizontal \( (x, y) \) direction (14).

In the longitudinal plane, the vertical strain, \( \varepsilon_{zz} \) (I), and the horizontal strain, \( \varepsilon_{yy} \), are measured. The vertical strain \( [\varepsilon_{zz} \text{ (I)}] \) is the strain orthogonal to the arm and parallel to the US probe, whereas the horizontal strain \( (\varepsilon_{xx}) \) is the strain orthogonal to the arm and orthogonal to the US probe. It may be noted that along the central line, theoretically, \( \varepsilon_{xx} \) (I) and \( \varepsilon_{yy} \) (II) should be equal.

**Statistics.** The mean strain curves as a function of time for the m. biceps brachii were calculated, as well as the strain values during maximum contraction within the entire m. biceps brachii. The intra-subject and intersubject variability of the average maximum strain values was calculated. The Pearson’s linear correlation coefficient was used to assess the similarity between the twitch contraction curves of the three acquisitions for all volunteers. The increase and decay in both force and strain curves were determined as the mean half-time, and the Student’s \( t \)-test was applied to determine possible significant differences between force and strain half-times. Furthermore, the Student’s \( t \)-test was used for comparison of the maximum vertical strains, individually measured in both planes, during twitch contraction. For the voluntary contractions, linear regression analysis was performed on the vertical strain estimates (from both planes) during the plateau phase vs. the measured MVC using a 95% confidence interval.

**RESULTS**

Examples of resulting strain data for a single-twitch contraction are shown in Fig. 5. For a large region within the m. biceps brachii (Fig. 5A), the strain curves are given (Fig. 5B). The strain curves show the global deformation behavior of the muscle as a function of time. The strain images visualize the local strain distribution within the muscle at several moments in time (Fig. 5C–F). The longitudinal images of the m. biceps brachii
reveal that the maximum vertical strain is in the center of the muscle. Vertical strain values up to 50% were measured. The cross-sectional view reveals different local regions of high strain. The horizontal strain reveals shortening up to \( \frac{\theta}{20\%} \) in the m. biceps brachii. The horizontal strain images, \( \varepsilon_{yy} \) and \( \varepsilon_{xx} \), have a noisier appearance than \( \varepsilon_{zz} \) (Fig. 5 E–F).

The plots in Fig. 7 show the mean values of the vertical and horizontal strain components during the phase of maximum strain during twitch contraction within the m. biceps brachii for all five subjects. For all subjects, the intrasubject variability of the vertical strains was found to be 3.1% \( (\varepsilon_{zz}(I)) \) and 2.1% \( (\varepsilon_{zz}(II)) \). The intersubject vertical strain was \( \varepsilon_{zz}(I) = 19.7 \pm 6.9\% \) and \( \varepsilon_{zz}(II) = 18.8 \pm 5.4\% \). A Student’s \( t \)-test revealed no significant difference in \( \varepsilon_{zz} \) during the phase of maximum contraction between the cross-sectional and longitudinal strain images \( (P < 0.12) \). The mean Fisher \( z \)-transformed correlation
of the vertical strain curves was 0.97 \( \varepsilon_{zz}(I) \), 95% confidence interval \( 0.59 \) – \( 1.0 \) and 0.97 \( \varepsilon_{zz}(II) \), 95% confidence interval \( 0.79 \) – \( 1.0 \) for all subjects. Because the correlation data were not normally distributed, the Fisher \( z \)-transformation was used to calculate the mean and 95% confidence interval of the correlation values. The intersubject horizontal strains and variability were estimated to be \( \varepsilon_{yy} / H_{11005} / H_{11002} \) \( 9.2 \) / \( 5.9 \)%, whereas the intrasubject variability was 3.5% (\( \varepsilon_{yy} \)) and 1.8% (\( \varepsilon_{xx} \)). For the horizontal strain curves, the correlations were 0.66 (\( \varepsilon_{yy} \), 95% confidence interval \( 0.67 \) – \( 0.98 \)) and 0.86 (\( \varepsilon_{xx} \), 95% confidence interval \( 0.63 \) – \( 1.0 \)), revealing a lower precision of the horizontal strain estimates.

Results for all of the volunteers reveal that the strain increases approximately at the same rate (mean half-time, \( \tau_{1/2,\text{mean}} = 18.4 \pm 23.1 \) ms) as the measured force (\( \tau_{1/2,\text{mean}} = 18.5 \pm 4.1 \) ms) during contraction (Student’s \( t \)-test, \( P < 0.02 \)). However, the mean force curve decays considerably faster (mean half-time, \( \tau_{1/2,\text{mean}} = 69 \pm 24 \) ms) than the vertical strain (\( \tau_{1/2,\text{mean}} = 214 \pm 161 \) ms). The difference was found to be significant (Student’s \( t \)-test, \( P < 0.01 \)).

The comparison of the three different types of electric nerve stimulation is illustrated in Fig. 8. The strain curves in all four directions are shown for single-pulse stimulation, a pulse train at 10 Hz and for pulse train stimulation at 20 Hz for one volunteer (see DISCUSSION). At 10 Hz, the force profile dropped slightly prior to the next pulse, so that an alternating wave shape profile becomes visible (Fig. 8E). The strain curves nicely follow this pattern (Fig. 8, A–D). At 20 Hz, this profile is no longer visible as the next pulse arrives before the relaxation phase starts. The mean strains at 20 Hz are \( \sim 30 – 50 \)% higher compared with the 10 Hz pulse strain. No distinct strain peaks related to the subsequent pulses can be found in the curves, which is in accordance with the measured smooth developing force profile.

An example of the voluntary contraction experiment is shown in Fig. 9. The measured force (Fig. 9A) and vertical strain curves (Fig. 9B) show a good resemblance. The mean vertical strain is already relatively high for 30% of the MVC, and the increase is moderate for higher force levels (Fig. 9B). The strain images captured during the plateau phase show an increase in strain and different patterns for increasing force (Fig. 9C). The measured mean vertical strain estimates (from both planes) during the plateau phase are also shown as a function of the exerted force (Fig. 9D). Results of linear regression are provided (\( R = 0.75 \)), including the 95% confidence intervals. Strain increases with force as expected.

DISCUSSION

We presented a method to measure strain in contracting skeletal muscles using biplane ultrasound. The use of biplane
Innovative Methodology

ultrasound enabled dynamic imaging and estimation of local muscle deformation in multiple directions for different types of induced muscle contraction patterns.

Using the proposed method, we measured 2D strain locally within two orthogonal planes in skeletal muscle, revealing the deformation in both the cross-sectional and longitudinal direction. The 2D displacement algorithm was able to properly track the tissue displacement and estimate the strain using the RF-data, despite the high strain (and thus strain rates) that occurred during muscle contraction and despite the observed rotation of the muscle tissue. Because of the lower sampling density in the lateral direction and the lack of adequate phase information in that direction, the horizontal strain images and curves are of lower quality. This is corroborated by the measured range of strain values and the correlation and similarity of the strain curves. Below, we discuss the feasibility of the biplane method for each of the different muscle contraction types: 1) The twitch experiments led to a controlled contraction and were used to demonstrate the reproducibility of the strain measurements. The maximum strain values showed a fair reproducibility. The shape of the strain curves also showed good similarity in all three directions for each subject (see RESULTS). The observed range of strains during twitch contraction was already quite high (10–30%), thereby already showing the need for a technique that is capable of measuring high strains. This induces the need of using a coarse-to-fine algorithm. Large signal windows are required to track the large global motion, whereas small signal windows are required when high strain is present. The muscle data contain both features. 2) The pulse train example nicely illustrates the performance of our technique in following a submaximal-tetanic contraction. The shape of the strain curves clearly revealed the difference between the different stimuli. However, electrical nerve stimulation with a train of pulses is not very comfortable, and in the m. biceps brachii, it is not easily performed. The induced contraction after a train of pulses resulted into relative large motion of the entire arm, moving away from both the stimulation electrodes and the ultrasound transducer (even though it was fixated). This movement is not expected to have affected the results substantially, but obtaining the reproducibility of the stimulus train was not trivial. 3) The feasibility and reproducibility of measuring strain during voluntary contraction were also examined. The reproducibility appears to be better for the lower force values (Fig. 10), which can be explained by the fact that lower strains can be measured more accurately. However, the tracking method was able to trace the strain at 60% and 100% MVC despite the fast contraction and relaxation of the muscle. Hence, this study reveals that strain measurements during voluntary contraction are indeed possible. Further research is needed to determine the relation between muscle excitation, force, and strain in more detail. For instance, the vertical strain at 30% MVC was relatively high (Fig. 9) and increased moderately for higher MVC values. This might be explained by opposing forces, caused by the stiffness of the surrounding connective tissue. In addition, other parameters could be of interest, such as the average strain rate during contraction, or the relaxation time.

Intrasubject and intersubject reproducibility is not easily quantified, considering the wide range of strain values during contraction and the noise level. However, the intrasubject and intersubject variability and the similarity of the strain curves (see RESULTS and Figs. 6 and 7) indicate that the measurements are consistent between (and within) subjects. Especially the resemblance between the vertical strain in the longitudinal and in the cross-sectional plane is promising (Fig. 6, A and B, 7A). The intrasubject variability (1.8–3.5%) was found to be lower than the intersubject variability (3.1–6.9%), revealing that the largest variance exists between the different subjects but with a good reproducibility for each individual subject. Although the force and strain curves of the twitch experiment showed a high resemblance, no distinct relationship between force and strain was found for this experiment. This was not the case for the voluntary contraction experiment, where a linear correlation between force and strain was found (Fig. 10), and each subject could serve as his own reference. However, the high strains observed at 30% MVC suggest that force and strain are not directly proportional.

We found that relaxation of muscle tissue measured with strain is significantly prolonged compared with the reduction of force in all different types of muscle contraction (Figs. 6, 8, and 9). The motion was also visible in the M-mode image (data not shown). A possible explanation may be that the subject slightly contracted the muscle immediately after the stimulus or that a delay between force and movement is present. Measurement errors, such as the accumulation of tracking errors, use of a large ROI, and out-of-plane motion, could have led to an overestimation of this finding. However, this seems unlikely, since the five stimulus pulses (10 Hz) were followed accurately by the strain estimation algorithm.

Other techniques may be applied also for deformation measurements. Conventional ultrasound B-mode data can be analyzed by 2D speckle tracking techniques to obtain 2D strain.
The advantage of B-mode speckle tracking is its robustness. However, the resolution of B-mode data is much lower compared with RF data, and the sensitivity of RF-based strain imaging is higher in the axial direction and approximately equal in the lateral direction (18). Secondly, the use of three-dimensional techniques should be considered. Full 3D volume imaging is currently available, including 3D RF-data (13, 17). Obviously, 3D data would yield the full 3D strain tensor within 3D space, which is necessary to characterize the heterogeneous and anisotropic properties of the muscle locally. However, sufficient temporal resolution is required and can only be obtained at the cost of a limited image view angle or of spatial resolution. This is not an attractive option when examining skeletal muscle contraction, since multiple (triggered) contractions are required to obtain RF data at a sufficient frame rate when using a standard 3D system. For instance, 3D cardiac data are normally acquired using ECG triggering in four to seven consecutive heart beats, resulting in a frame rate of 35–50 Hz. Since fatigue may be induced and reproducibility is not high enough for voluntary contractions, this acquisition scheme is not preferred. Moreover, 3D imaging will increase both experiment time (and patient’s burden) and computational load. Finally, the field-of-view of ultrasound data is limited compared with MRI (27). However, ultrasound data acquisition has the primary advantages of being cost-effective as well as its ease of use. Another benefit to point out is that the temporal and spatial resolution is higher compared with MRI. Besides, the so-called “speckle” appearance of the ultrasound images enables displacement estimation without the use of tagging.

The vertical and horizontal strain images have shown the advantage of displaying local information of the strain, whereas mean strain curves only reveal the global deformation of the tissue and are limited to one dimension [comparable to force curves and the work on aponeurosis strain measurements (22)]. The fact that strain differs locally can be seen in the strain images (Figs. 5 and 9) during maximum contraction. Such nonuniform strain distributions have been reported also by others. The experimental MRI findings of Pappas et al. (27) showed a homogeneous strain distribution over the anterior fascicles, but a heterogeneous deformation pattern along the centerline of the m. biceps brachii. It was hypothesized that this heterogeneity might be caused by the presence of the aponeurosis and tendons and the difference in stiffness of these two with respect to the muscle tissue. These experimental findings were corroborated by the simulation work of Blemker et al. (2). However, the field-of-view of the biplane ultrasound images is too small to measure the strain variation over the entire muscle as performed by Pappas et al. (27). Besides, in those experiments, a lower MVC of 15% was used and muscle shortening was examined in the fiber direction but not in the orthogonal direction (equal to vertical strain), which makes a direct comparison with our results difficult. We have shown that our...
The clinical applications of measuring local skeletal muscle strain are diverse. Measurement of the time and amount of strain in relation to force could be of use in studies regarding force transmission. This could be of special interest in disorders of connective tissue such as Ehlers-Danlos Syndrome. Also force generation in primary muscle disorders such as muscular dystrophies could be studied using strain-force patterns. As strain appears to diminish more slowly than force, it might be a more sensitive method to detect disorders with delayed relaxation such as myotonic disorders. Using the presented method, one can assess local multidimensional strains and strain differences within a muscle. This can be of interest in disorders with focal muscle involvement, such as inflammatory muscle diseases (30). Also disorders affecting peripheral nerves would be interesting to study using skeletal muscle strain. However, further research is needed to investigate whether skeletal muscle strain measurement can detect differences on a motor unit level, as this could be used to monitor disease progression such as amyotrophic lateral sclerosis (1).
Conclusion. In conclusion, we introduced a multidimensional strain method using biplane imaging in skeletal muscles during induced and voluntary contraction. The method revealed good performance using RF data in tracking the tissue motion and enabled the measurement of local deformation, even despite the large displacement and strain between the subsequently acquired echo frames and despite the moderate frame rate of the ultrasound equipment in biplane mode.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the authors.

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