Concurrent assessment of individual motoneuron discharges and muscle unit displacement velocity in humans by high-density EMG and ultrafast ultrasound

Running title: Characterising the motor unit electrically and mechanically *in vivo*

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Key Words

electromyogram; ultrasound; ultrafast ultrasound; motor unit; skeletal muscle; excitation-contraction coupling.

Key points

- We identified the activity of single motor units (MUs) from high-density surface electromyography (HDsEMG) and used this information in combination with ultrafast ultrasound to extract local muscle motion due to the contraction of individual muscle units
- Multiple MUs, including those with fibres overlapping in space, can be simultaneously and individually detected using this technique
- The proposed method allows us to measure both the spiking activity of motor units and their movement within the muscles concomitantly
- The technique allows for populations of MUs to be tracked and monitored in the electrical and mechanical domains simultaneously and non-invasively during natural contractions, thus achieving a high spatio-temporal resolution in the characterization of MU behaviour

Abstract

The study of human neuromechanical control at the motor unit (MU) level has predominantly focussed on electrical activity and force generation, whilst the link between these, the muscle deformation, has not been widely studied. Here, we describe a methodology utilising ultrafast ultrasound (US), allowing imaging of up to tens of thousands of frames per second, to measure the deformation of the muscular tissue due to individual MU twitches for a population of active MUs during voluntary contractions. We used the spiking activity of MUs decomposed from high-density surface electromyography recordings of the tibialis anterior muscle to guide the analysis of simultaneously recorded ultrafast US. With a novel analysis on the US images we identified, with high spatio-temporal precision, the velocity maps associated with single MU movements. From the individual MU profiles obtained from the velocity maps, the region of movement, the duration of the mechanical twitch, the total and active contraction times, and the activation time were computed. The latter features, the temporal features, showed high repeatability across different force levels. The former feature, the spatial feature, showed high consistency across force levels, however the complicated dynamics of the muscle motion resulted in morphing and translation of these regions. Furthermore, the experimental measures provided the first evidence of muscle unit twisting during voluntary contractions. The proposed approach allows, for the first time, non-invasive recordings of muscle deformation due to individual MU activations during voluntary contractions.

Introduction

Motor units (MUs) translate synaptic input into muscle motion: the action potentials (APs) discharged by the spinal motoneurons (MNs) in response to synaptic input evoke transient mechanical contractions, called twitches, of the fibres of the muscle unit. Ultimately, global motion is a result of the summation of these muscle unit twitches. The neuromechanics of movement can therefore be studied at the MU level, where the ultimate neural code of movement is translated into mechanical output.

Methods for MU characterisation can be divided into two broad categories – those which consider the electrical activity, and those which consider the mechanical output. The electrical behaviour of MNs is usually analysed with electromyography (EMG), either intramuscular (iEMG) or surface (sEMG). Both iEMG and sEMG allow the indirect

identification of MN discharges by the analysis of the corresponding muscle unit APs. On the other hand, the mechanical response of entire muscles can be studied using ultrasound (US) (Botter *et al.*, 2013; Dieterich *et al.*, 2017; Tweedell *et al.*, 2019). Whilst US has also been used for studying the mechanical response of individual MUs, these approaches are limited to electrically stimulated contractions (Deffieux *et al.*, 2008; Waasdorp *et al.*, 2019, 2021), or by their reliance on invasive EMG techniques (Rohlén *et al.*, 2020*b*, 2020*a*).

Human neuromechanical control is well explored from both an electrical perspective, providing little spatial detail, and in terms of muscle contraction, mainly treating the muscle as a global force producer. However, how the microstructures in the muscle combine to do so is less clear – the complexity and linearity of the system are unknown. These are important considerations for understanding strategies of human muscle control. A representative example of a mechanism relevant for understanding the neuromechanics of muscle control is excitation-contraction coupling (E-C coupling), which is one of the components of the electromechanical delay (EMD). E-C coupling refers to the communication between the electrical activity in the skeletal muscle fibre membrane and the release of calcium, eliciting the contraction (Calderón *et al.*, 2014). The ability to measure E-C coupling in-vivo in voluntary contractions for individual MUs would provide insights into neural control as the process of modulating MU activities, and would also enable detection and monitoring of pathological conditions in which the E-C coupling is affected, such as myotonic dystrophy 1 (Esposito *et al.*, 2016) and lower back pain (Vasseljen *et al.*, 2006). Coupling time has been measured in electrically stimulated contractions (Nordez *et al.*, 2009; Hug *et al.*, 2011*a*; Esposito *et al.*, 2011) or in voluntary contractions at the global muscle level with problematic reliance on bipolar EMG (Hug *et al.*, 2011*b*; Begovic *et al.*, 2014).

Here, we present a non-invasive recording and processing approach that accurately identifies individual MN discharge times and muscle unit movements during voluntary contractions. The proposed approach is based on pairing real-time high-density sEMG (HDsEMG) decomposition, biofeedback, ultrafast US imaging, and a novel spike-triggered average (STA) processing method. We describe the proposed methodology, we test it by comparing some of the extracted physiological parameters with literature data, and we provide results on new physiological measurements not previously possible, including MU-specific E-C coupling time, allowing deeper insight into neuromechanics.

Methods

Ethical approval

All procedures and experiments were approved by the Imperial College Research Ethics Committee (ICREC reference: 20IC6422) in accordance with the declaration of Helsinki. As the datasets acquired are large, the data was not registered in a public database but may be made available, as appropriate and reasonable, upon request.

Experimental Setup and Equipment

The method described is based on synchronised acquisition of ultrafast US, HDsEMG, and force data during repeated sustained contractions performed with feedback on MU activity provided by real-time HDsEMG decomposition. Fig 1. summarises how APs generate contractions and shows the information captured by each modality.

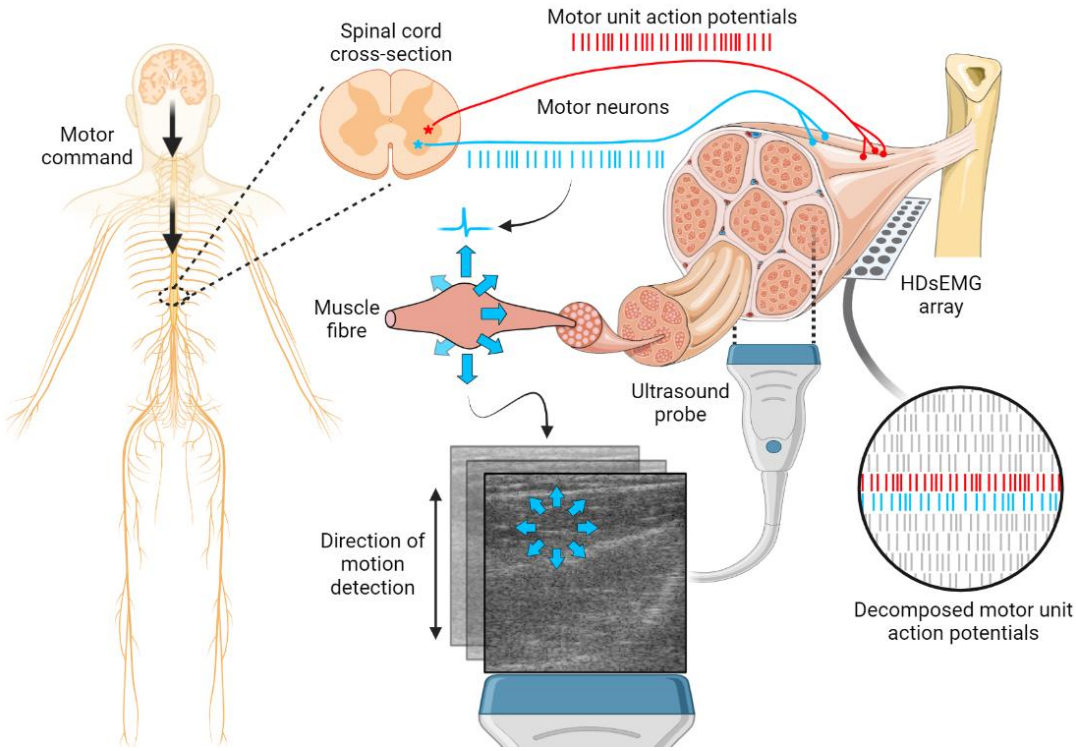

Figure 1. Diagram of the experimental set-up and the underlying human neuromechanical control mechanisms. Motor command is passed from the brain via the spinal cord to the motoneurons (MNs). Each MN innervates a group of muscle fibres. The ensemble of action potential (AP) trains from the pool of MNs innervating a muscle are often referred to as the neural drive to the muscle. Together, the innervated muscle fibres (muscle unit) and the MN constitute a motor unit (MU). APs travel along the MN axon and generate

corresponding APs along the muscle fibre membranes. These cause a contraction (shortening and thickening) of the fibres. The ultrasound (US) imaging plane is perpendicular to the direction of the fibres, so a radial expansion is seen in the imaging plane. The velocity mapping algorithm detects axial motion only, so only the components of the expansion perpendicular to the probe are detected. High-density surface EMG (HDsEMG) arrays (for clarity, only one is shown in this diagram) are placed over the surface of the skin, and an online decomposition algorithm is used to estimate the motor unit action potential (MUAP) discharge times.

Ultrafast US

The US data were acquired with the Vantage Research Ultrasound Platform (Verasonics Vantage 256, Kirkland, WA, USA), equipped with a L11-4v transducer with 128 elements and centre frequency of 7.24 MHz. Single angle plane wave isonification was used to enable a frame rate of 1000 frames per second with acquisitions of 30 seconds, resulting in 30,000 frames of US data per recording. The data was beamformed using delay and sum beamforming, resulting in images of 357 pixels axially and 128 laterally, corresponding to pixel dimensions of 0.1 and 0.3 mm. The frame rate and recording duration were chosen as per memory limitations. The chosen values proved to be a good compromise between temporal resolution and duration for accurate processing. The acquisition and control were performed by custom codes written in MATLAB (Matworks, Massachusetts, USA).

High-Density surface Electromyography

The HDsEMG signals were recorded by two electrode arrays of 64 channels (5 columns and 13 rows; gold coated; 1 mm diameter; 8 mm interelectrode distance; OT Bioelecttronica, Torino, Italy). The signals were recorded in monopolar derivation, amplified, sampled at 2048 Hz, A/D converted to 16 bits with gain 150, and digitally bandpass filtered (10-500 Hz), using an EMG pre-amp and a Quattrocento Amplifier (OT Bioelettronica, Torino, Italy).

Force

An ankle dynamometer was used to measure the dorsiflexion force. The leg and foot were constrained in the dynamometer using foam padding and secured using straps to provide stability and to eliminate other motions, keeping the foot at an angle of 90 degrees to the leg. The force data was fed through a Forza force amplifier (OT Bioelecttronica, Torino, Italy) directly into the auxiliary port of the Quattrocento Amplifier.

Synchronisation

A 1 µs active low output was fed from the Verasonics US system into an Arduino UNO that in turn outputs a 490 µs pulse, elongated to allow for detection at the 2048 Hz sampling frequency of the EMG Amplifier. The trigger was recorded at the start and end of each recording period and used to align the data in post-processing. To test the accuracy of synchronisation and alignment of the US series and the HDsEMG signal, the HDsEMG grid was sharply perturbed such that a peak was seen in the electrical (as artefact) and velocity signal obtained from the US series. These peaks were seen to align within the expected error of ± 0.5 ms resulting from the US frame rate, confirming the validity of the synchronisation method.

Participants

A total of 12 healthy participants (mean age, 26.25 ± 2.8, 4 female), were recruited for this study. Before the experiment, the volunteers were fully briefed on the experiment, provided with a participant information form, and asked to sign an informed consent form.

Setup

The experiment was conducted on the tibialis anterior (TA) muscle. The motivation for using the TA in this study was threefold: it is a large, long muscle allowing space for all the sensors used; it is superficial, resulting in high-quality HDsEMG decomposition (Del Vecchio *et al.*, 2020); and its fibres are long and have a relatively low pennation angle (Sopher *et al.*, 2017). The latter property allowed for mounting the ultrasound probe and the HDsEMG arrays in separate areas whilst ensuring they still overlay the same fibres, and ensuring more uniform velocity fields within the imaging plane. In all participants, the experiment was performed on the (dominant) right leg.

Before beginning the experiment, the skin over the TA was shaved, lightly treated with a chemical abrasive, and cleansed using an alcohol spray. The muscle belly was then identified by palpation whilst the participant was guided to contract and relax the TA. A HDsEMG electrode array was placed over the proximal part of the muscle, parallel and lateral to the tibia. The grid was positioned such that all electrodes were on top of the TA belly. A 1.5 cm gap was left distally with respect to the first array and a second array was placed over the distal part of the muscle, also parallel and lateral to the tibia. In some participants, the distal electrodes of the second array fell outside the TA belly, which resulted in small signals and did not impact the results. The electrodes were placed along the estimated direction of the

muscle fibres and secured using Tegaderm Film Dressings and medical self-adhesive bandages.

Next, the US probe was placed in the gap between the EMG electrodes, with the imaging plane perpendicular to the length of the muscle fibres. A water-based US gel was used to improve the coupling between the probe and the skin. The US probe was held by a custom-built probe holder and secured in position using straps. The participant sat in a chair in a comfortable position with the leg strapped into the ankle dynamometer. The US system was turned on to perform real-time imaging and small adjustments were made to ensure the TA muscle belly was covered. Finally, a computer screen was adjusted to a comfortable distance and position to provide visual feedback during the experiment. The described configuration of separate locations for the EMG grids and US probe was preferred over the possibility of using grids transparent to US directly under the US probe (Botter *et al.*, 2013) since transparent grids have not been tested for single muscle unit motions (see also Discussion).

Experiments

Real-time online decomposition of the HDsEMG signals into MU discharge times was performed using the methodology of Barsakcioglu et al. (Barsakcioglu *et al.*, 2021). The identified spike trains were provided as visual feedback to the participant throughout the experiment. This approach ensured continuous activation of the target MUs, and allowed the participant to precisely control the number of recruited MUs.

The online decomposition was trained on a contraction at 10% MVC, maintained for 41 s. Once the online decomposition model was trained, the subject was asked to slowly increase the dorsiflexion force whilst the experimenter inspected the decomposition to determine whether the model was satisfactorily trained, and the noise level was acceptable. Model training was repeated when deemed necessary (this was never done more than once during the experiments).

With the decomposition trained, the visual feedback displayed the MU discharge times in real-time. The subject was asked to slowly increase the dorsiflexion force, thereby slowly increasing the number of recruited MUs displayed on the screen. The order of recruitment of the identified MUs was noted. The subject was then trained to recruit individual MUs and to incrementally increase the number of MUs in a controlled manner. Providing visual feedback was a vital part of the training as fine recruitment of low threshold MUs is a complex task.

Once the participant was comfortable with this control method, the recordings commenced. The subject was instructed to recruit a single MU and to keep it active for 30 s whilst the US and HDsEMG recordings were taken. The subject then rested for 60 s before repeating this process for the same MU two further times. Each 30 s paired recording of HDsEMG and ultrafast US was considered a single trial. The three trials were then repeated for up to six different MUs. If fewer than 6 MUs were identified, the process was only performed up to the maximum number of identified MUs.

Data Analysis and Processing

Datasets

For one subject the synchronisation signal failed, and for another the HDsEMG decomposition failed, resulting in valid data from 10 subjects. The final dataset (Table 1) included 148 paired recordings.

Participant	MUs Detected	Quality MUs	Total Paired Recordings	Quality Paired Recordings	MU-STA curves obtained
1	8	6	18	12	33
2	-	-	-	-	-
3	7	7	19	14	50
4	5	3	18	12	19
5	6	5	19	17	51
6	-	-	-	-	-
7	4	4	14	12	31
8	7	5	18	17	39
9	7	5	18	14	40
10	7	4	18	18	28
11	9	6	17	17	57
12	8	6	20	15	42
Total	68	51	179	148	390

Table 1. Participant and recording information. "MUs detected" is the total number of motor units (MUs) that were recorded using EMG at any time during an ultrasound (US) recording. "Quality MUs" is the total number of MUs that were properly visible in the processed US recordings and of sufficient quality for identification. "Total Paired recordings" is the total number of paired recordings made for each subject. "Quality Paired Recordings" is the total number of recordings with at least one MU of sufficient quality i.e. the number of recordings used in data analysis. "MU-STA curves obtained" is the total number of motor unit spike-triggered average (MU-STA) curves produced for each participant (see Data Processing).

Data Processing

From the online decomposition, the discharge times of the identified MUs were available. However, each discharge was identified with some variability with respect to the AP waveform. Since in the current application it was important to define the precise onset of the MU discharge, we estimated the required temporal shift so all APs would be aligned in time. The shift was identified by means of a STA of the sEMG signals over each channel, producing the average AP shape for each MU. A double differential AP shape was calculated and the channel with highest signal-to-noise ratio was selected, and the onset of the action potential was defined as the first time sample with amplitude higher than 5 standard deviations of the baseline signal (Ibáñez *et al.*, 2021). The time difference between the onset and the centre of the STA window was the shift by which all AP spikes of the MU were shifted, such that the corrected AP discharge time reflected the onset of electrical potential at the neuromuscular junction in a consistent manner for all discharges.

For the ultrafast US data, each trial comprised 30,000 frames (357 x 128 pixels each) that were used to generate a series of velocity maps using 2D autocorrelation velocity tracking (Loupas *et al.*, 1995) with a sliding window of 2 frames, corresponding to 2 ms. This resulted in 29998 frames (357 x 128 pixels) with bipolar amplitude representing axial velocity (negative pointing away from the probe). The recordings were then further low-pass filtered (0.5 MHz) along the axial direction and band-pass filtered (5 – 100 Hz) along the temporal direction, removing noise from outside the expected temporal range of the response.

Next, a STA of the US velocity maps was performed using the discharge times identified from the HDsEMG decomposition. For each identified MU, a $\pm$50 ms window (100 frames in total) of velocity data around each discharge was selected (Fig. 2). All windows were averaged, resulting in 100 frames of STA US maps (US-STA maps) for each MU (Fig. 3a).

Supplementary Video 1 in the Supporting Information shows an example of the 100 frame US-STA i.e. the average muscular movement caused by an individual MU discharge. In this representative video, a large part of the movement caused by the target MU is hidden by the motion of non-muscular structures. These high noise areas need to be removed for an accurate analysis of the muscle unit movement. For this purpose, for each US-STA window, the variance across all windows was calculated (Fig. 3b). To determine the direction of motion of a pixel as induced by the MU activity (either upwards or downwards), we performed a subtraction of the sum of the second half of the STA from the first (Fig. 3c). For each pixel the sum over the squared STA divided by the variance (with the convention of a

positive sign for upwards and negative sign for downwards motions) was calculated, highlighting regions of intense (high amplitude) and consistent (low variance) movement, whilst penalising regions of non-consistent movement (e.g., tibia movement). Lastly, the map was downsampled axially by a factor of 3 (119 x 128 pixels wide; 0.3 x 0.3 mm per pixel) and the regions of no interest (skin and fat layers and the rows after the maximum sEMG range) were removed. Each of the final maps, which we will term MU activity maps (MUAMs) (55 x 128 pixels wide), represents the muscular movement caused by an individual MU discharge during a voluntary isometric contraction.

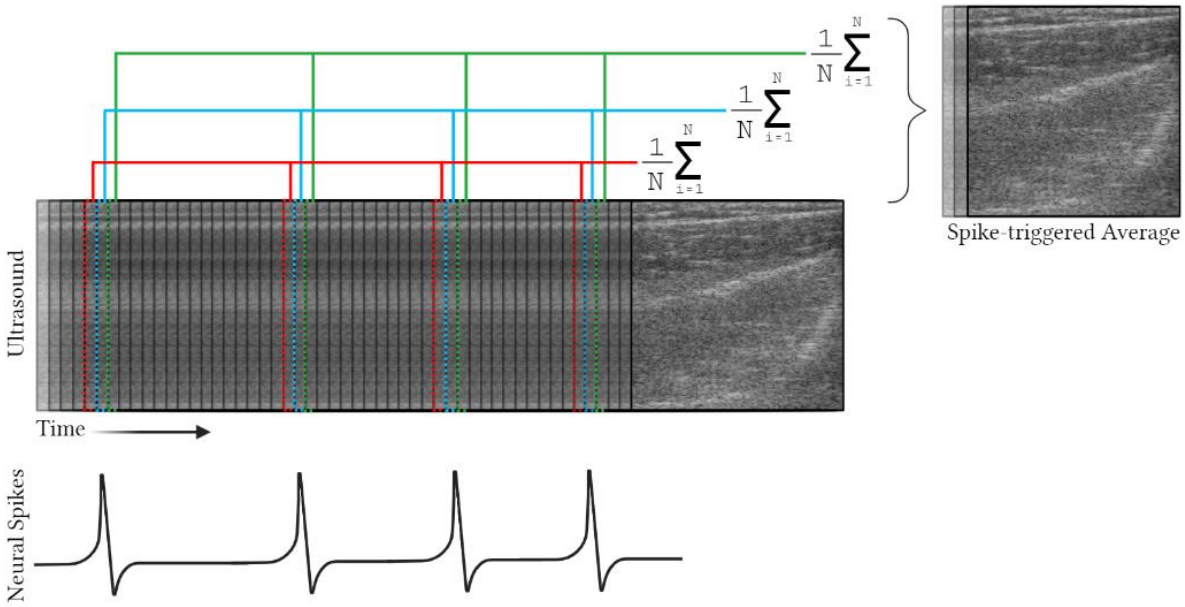

Figure 2. Diagram of the spike-triggered average (STA) method employed. For each motor unit (MU) and identified discharge, a window around the discharge timing is selected. All selected windows are summed and averaged, resulting in a single STA window for each MU. Note that this process is performed on the velocity fields obtained from the ultrasound image series.

Muscle Unit Area of Motion Identification

The MUAMs show, for each pixel, the effect caused by the discharge of a specific MU, with high values (positive or negative) representing regions of large motion that are highly synchronised with the MU discharge, whilst low values represent regions with either small or highly desynchronised motion. In some situations, a MUAM may show separate regions of positive and negative velocity associated with a single muscle unit motion. In these cases, the two regions were treated separately but assigned to be part of the same MU. To properly define which pixels were associated to the movement of a muscle unit, a thresholding

approach was used. For each MUAM, pixels with absolute value greater than 65% of the maximum amplitude were considered as part of the region of movement due to the MU activity. We will refer to the pixels constituting this region as the MU motion domain. In contrast to the MU territory, defined as the cross-sectional area containing the muscle fibres of a MU, the MU motion domain is therefore defined as the region of muscle tissue that shows movement with amplitude and synchronicity over a predefined threshold when a MU discharge occurs. Whilst the MU territory refers to the MU anatomy, the MU motion domain refers to the mechanics of the muscle unit.

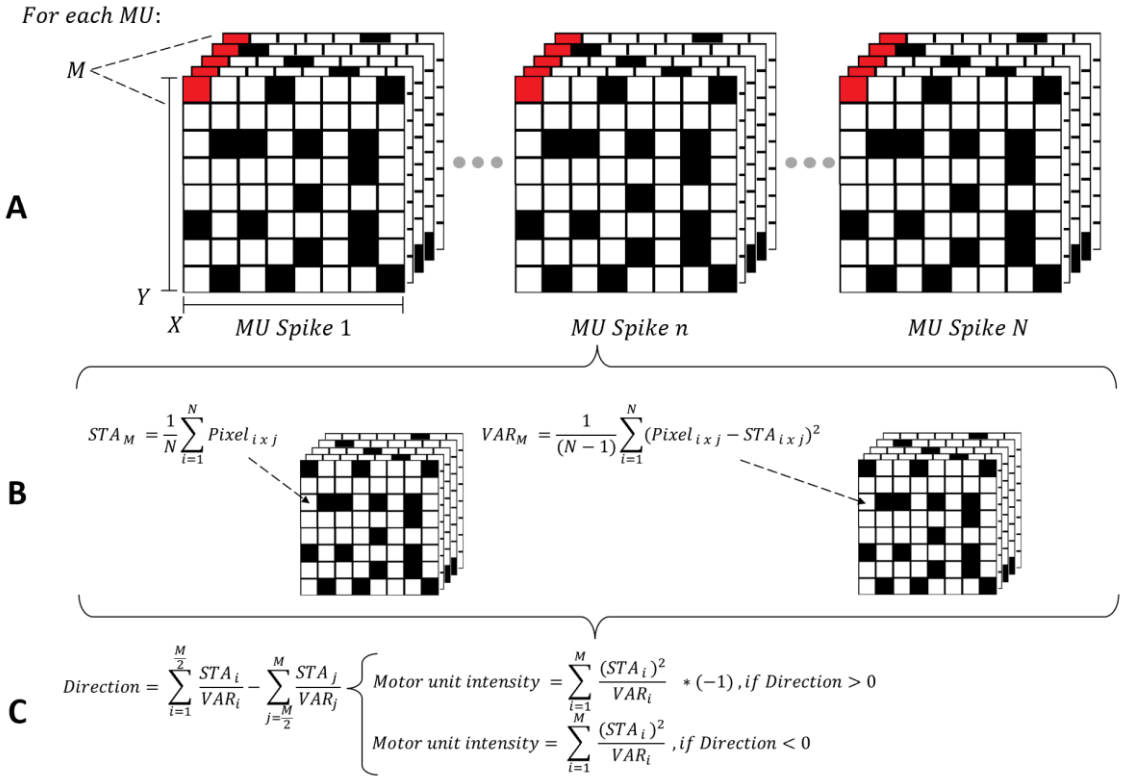

Figure 3. Process to highlight motor unit (MU) related motion. For each MU a window of 50 frames before and 50 frames after each action potential (AP) time is selected (shown in panel A, where M = 100 frames, and N = total number of APs in that recording) and a spike-triggered average (STA) technique is employed to obtain the STA and the variance of the STA, as shown in panel B. The STA is divided by its variance and summed over time to obtain a single map of intensities, where high intensity represents areas of high motion and low variability in the STA. The values are also multiplied by a direction factor (as shown in panel C) which separates segments of the STA map moving in opposite directions in response to the AP.

Motor Unit Mechanical Twitch Profile

The MUAMs were used with the US-STAs to generate a STA of the mechanical twitch of the MU (MU-STA curves). These were calculated by averaging the US-STA pixels curves inside the MU motion domain. One MU-STA curve twitch profile was generated for each MU (two

if the MU caused both a positive and negative movement in separate regions) in each trial. In this work, the term “twitch” always refers to the movement of the fibres in the muscle unit as measured in terms of displacement velocity.

Each extracted MU-STA curve is a unique twitch profile. These curves were analysed quantitatively for the extraction of physiological features of interest. The extracted parameters are shown in Fig. 4a. One extracted feature was the duration of the mechanical twitch – the time over which the tissue drastically changes its velocity profile. This corresponds to the time interval over which the tissue accelerates, thus the boundaries of this interval are inflection points. In some cases (Fig. 4b) the onset of the mechanical twitch can be easily identified by the inflection point of the velocity, however in other cases (Fig. 4c) the twitch occurs during a later stage of relaxation of the previous twitch, so although there is a sharp velocity change, the velocity inflection point cannot be used to define twitch onset. As such, the peak in the second derivative was used to mark the onset of the mechanical twitch.

Another feature of interest was the contraction time, defined as active or total. The active contraction time was defined as the interval between velocity zero crossing and the inflection point. The total contraction time starts at the same point, however it ends at the second zero crossing before the next spike. Although both features represent the time where the tissue is in the “contracting state”, the active contraction time accounts for the time over which the twitch is actively accelerating the tissue, whereas the total contraction time includes the full duration of the “contracting state” - the total contraction time includes both the time when the fibres are actively generating the contraction, as well as the time when the acceleration is due to the elastic recovery forces.

Finally, we also calculated the activation delay, defined here as the time between the onset of the electrical AP and the onset of the mechanical twitch. Currently, the electromechanical delay (EMD) is measured as the delay between the onset of the detected electrical activity and the onset of force production, and often its measure has significant inaccuracies due to the use of bipolar EMG at a distance from the motor point (Cavanagh & Komi, 1979). As such, in literature, the EMD includes the time lag due to the synaptic transmission, the excitation-contraction (E-C) coupling, the muscle force transmission on the series elastic components, the aponeurosis, and force transmission on the tendon. Our technique allows for the measurement of the delay between the onset of voluntary motor unit action potentials (MUAPs) and the first measurable acceleration on the fascicles of single muscle units.

Therefore, this measure includes the time lag caused by the E-C coupling and force transmission on the series elastic components (between the innervation point and the US measurement plane). However, as the force transmission travels at ~30 m/s (Morimoto & Takemori, 2007; Nordez *et al.*, 2009) and the end-plates for the detected MUs were within 4-cm distance to the US plane, the resulting delays were less than 1 ms. Thus, the measured activation delay mainly represented the E-C coupling, making our measurements the first to report individual MU E-C coupling delays.

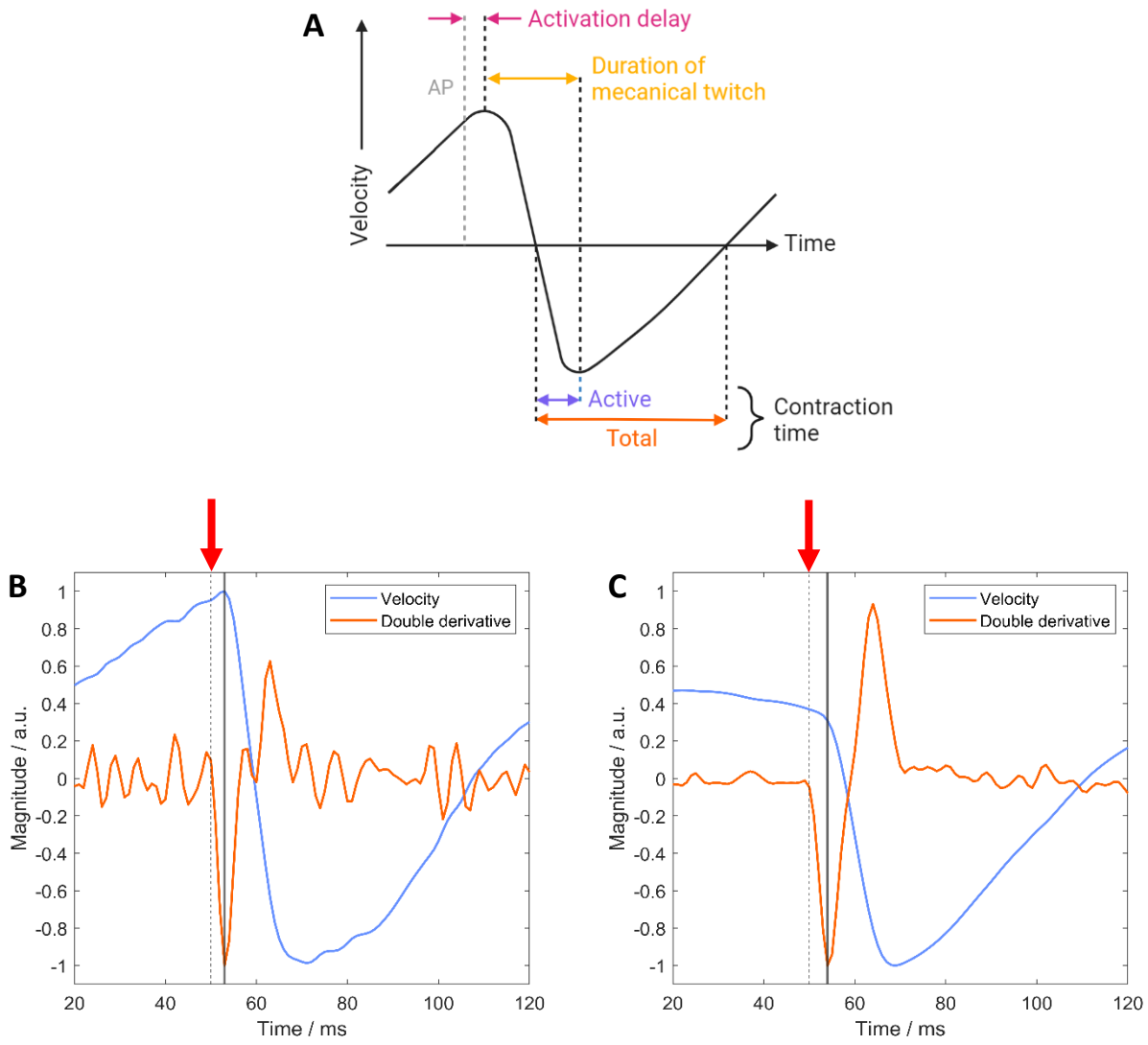

Figure 4. Measurements extracted from the motor unit spike-triggered average (MU-STA) curves. A) The dotted AP line is the time of the onset of the electrical action potential (AP) measured using high-density surface EMG. Note that before the effects of the next twitch begin, the fibre is still relaxing from the previous twitch, thus the velocity is non-zero and changing prior to the AP. Following the AP, the velocity reverses due to the twitch, and the time between the AP and the turning point is the activation delay. The fibre is then accelerating due to the twitch between the first and second inflection points – we refer to this as the duration of the

mechanical twitch. Between the two zero crossings, the fibre is in a contracted state – thus we break this region down into active contraction time, where the fibre is being actively contracted, and total contraction time, which also accounts for the relaxation back to the neutral position. B) An example velocity profile for a motor unit (MU) where the interspike interval is low, so an inflection point is seen at the onset of the mechanical twitch. C) An example velocity profile for a MU where the interspike interval is high, so the previous mechanical twitch is in a later stage of relaxation resulting in no turning point at the onset of the twitch. To enable twitch onset detection in both cases, the peak of the second derivative is used (solid black line). In each case, the dotted black line is the firing time from the EMG, highlighted by a red arrow above the plots.

Processed Dataset

Following the process described above, all 179 paired recordings were processed. However, 31 recordings were discarded as the HDsEMG decomposed spike trains were poorly identified. A total of 68 MUs were identified across all participants in the HDsEMG recordings, 51 of which (75%) could also be detected in the US recordings. This discrepancy is due to out of plane MU activity, a problem which is further explored in the Discussion. As such, 390 MU motion domains and mechanical twitch profiles from 51 MUs from 10 participants across 148 recordings were obtained (Table 1).

Statistical Analysis

The data presented in this paper are intended to act as a proof of concept, showing what the set-up and processing pipeline can achieve. For each value and graph, the number of data points, *n*, is stated in the accompanying text. For all histograms and box plots $n = 390$.

Results

From 12 participants we identified 51 MUs and, across all trials, we obtained 390 MUAMs and MU-STA curves (see Table 1). Fig. 5 shows an example of the two signal modalities plotted together, as well as their STAs. For each MU activation, the velocity inside the MU motion domain changes in a synchronous manner. Fig. 6 shows an example of the outputs for a trial with four identified MUs. The identified units have variable discharge rates (Fig. 6a), as is expected when the force level is close to the recruitment threshold of the MU. Fig. 6b and Fig. 6c compare the shapes of the electrical STA and the MU-STA – the electrical STA is unique to a given MU while the MU-STAs are similar in this example. The MUAMs in Fig. 6c are truncated at approximately 10 mm as throughout the experiment no MUs were

detected below this level, because of the relatively superficial detection volume of the surface EMG (Merletti & Muceli, 2019).

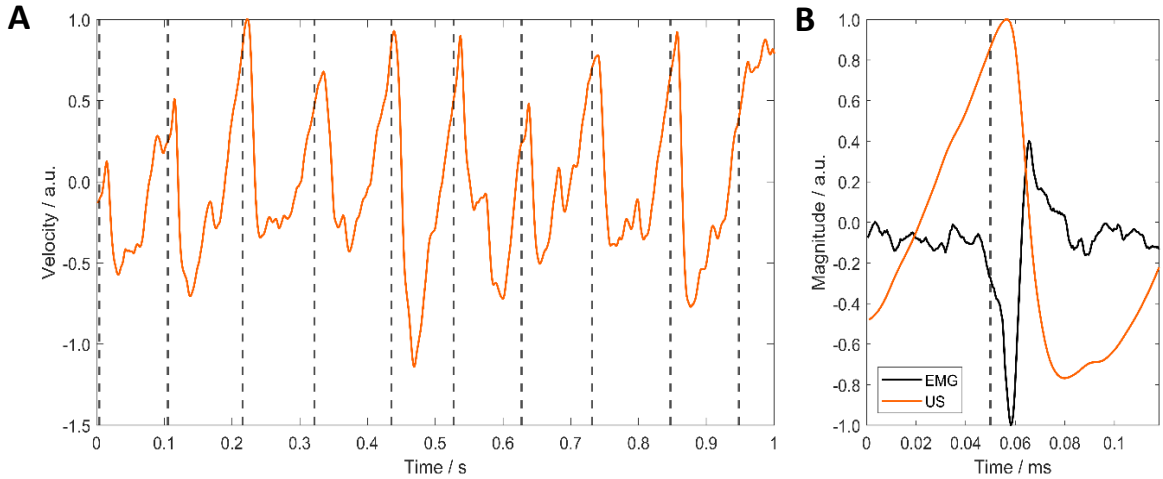

Figure 5. Synchronisation of velocity and EMG detected firing times. A) Example of average velocity inside motor unit (MU) motion domain for an identified MU (orange line), and action potential (AP) discharge times from EMG decomposition (black dotted lines) showing synchronicity. This is a 1 s interval extracted from a 30 s recording. B) Motor unit action potential (MUAP) found by means of spike-triggered average (STA) on the EMG signal (black line) and motor unit spike-triggered average (MU-STA) curve for the same MU (orange line). The black dotted line is the corrected firing time – the time at which the double differential of the highest signal to noise channel of the EMG array is above 5 standard deviations of the baseline noise level. The MU-STA is relaxing from the previous twitch hence the velocity is changing prior to the discharge time.

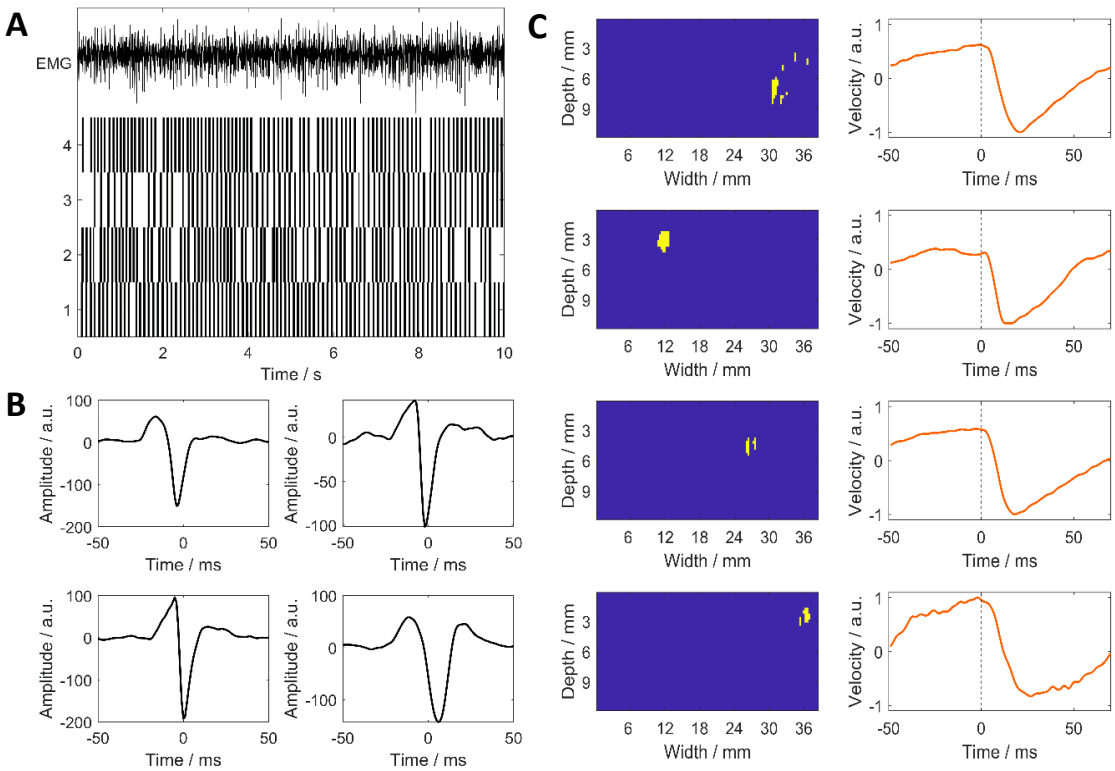

Figure 6. Example set of outcomes for a recording. A: Summed raw EMG signal (top) across the 64 recording electrodes in the high-density surface EMG array and the action potential (AP) discharge times of four of the identified motor units (MUs) – each of the units is given a label as shown to the left of the plot. A 10 s excerpt of the 30 s recording is shown. B: motor unit action potentials (MUAPs) for each of the identified MUs prior to correction to align activity onset, found by performing STA on the EMG signal. C: Motor unit activity maps (MUAMs) (left) and motor unit spike-triggered average (MU-STA) curves (right) for each of the four MUs. The MUAMs show the MU motion domains, and the MU-STAs show the average velocity in these regions in response to an AP. For all STA plots, the AP occurs at 0 ms.

Motor Unit Spatial Motion Domain

To analyse the consistency of the area assigned as the MU motion domain across repeated trials, the percentage overlap of the spatial domains for each MU across different trials was calculated. The results are shown in Fig. 7 for each participant, and the average for all participants was 47.0 $\pm$ 17.4 %. In contrast, the percentage overlap between a given MU motion domain and the motion domains of other MUs was 6.9 $\pm$ 6.8 %. This showed that MU motion domains were more similar to the motion domains of the same MU in different trials than to those of other MUs for all participants. Although a 47% of overlap across trials may seem a low value, it incorporates error resulting from thresholding, as well as morphing and shifting due to other active MUs. The consistency in location of MU domain can be seen for representative MUs in Fig. 8.

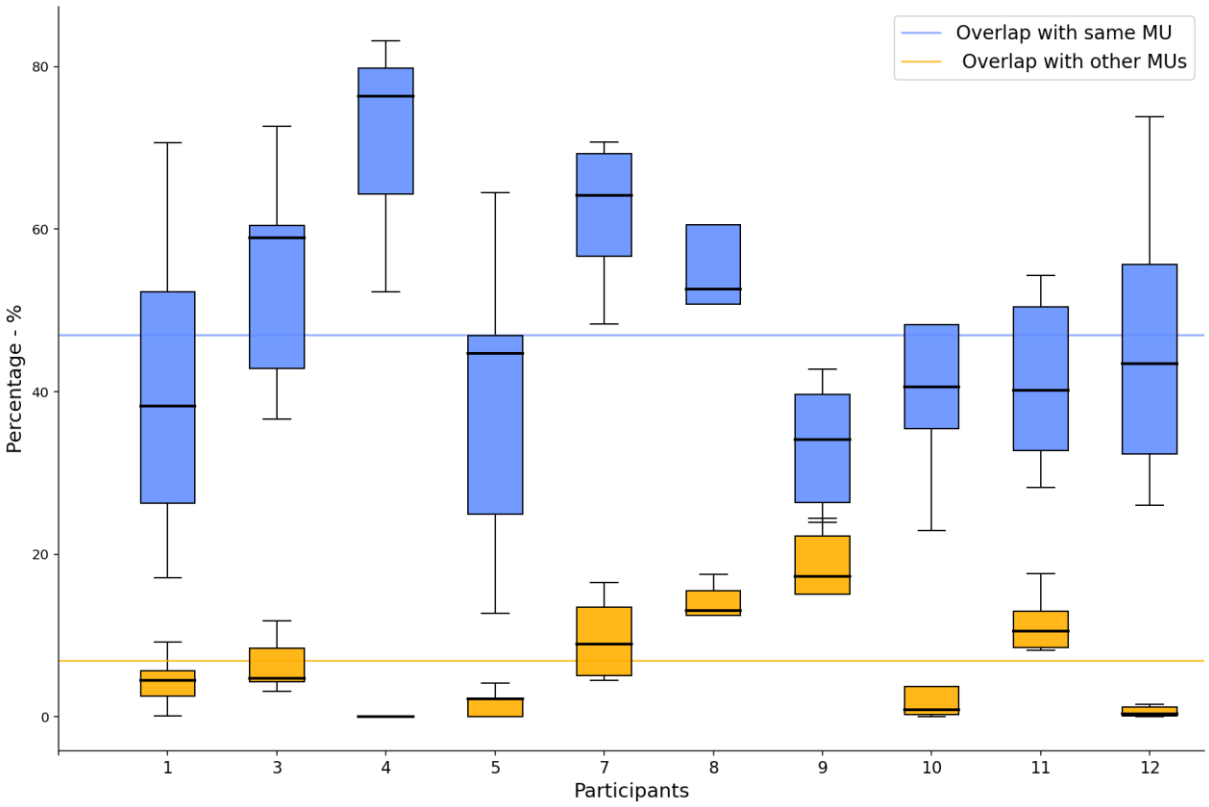

Figure 7. Box plot presenting participant-wise percentage overlaps of motor unit (MU) motion domains with other MU motion domains (orange), and with the same MU motion domains across different trials (blue). This is for 51 MUs identified in 390 trials across 10 subjects. For every participant the MU motion domains are more similar to the same MU motion domains in different trials than to other MU motion domains. This suggests that we are accurately identifying distinct MU motion domains from different simultaneously active units.

Mechanical Twitch Reproducibility

Fig. 8 shows, for two representative MUs, both the MUAMs (left panel) and the normalised MU-STA curves (right panel) obtained for different trials (Example MU 1: 17 trials, Example MU 2: 8 trials). Each grey curve is the MU-STA velocity profile, averaged over the MU motion domain. These curves show high consistency across trials, despite changes in conditions (differing number of active MUs). For all 51 MUs across the 390 trials, the average standard deviations between curves from the same unit was 6.69 %. From these results, we concluded that the MU-STA curves were highly repeatable across force levels. Whilst it is clear that the MU motion domain morphs and shifts at differing force levels, the twitch profile is consistent across force levels, suggesting higher linearity in the temporal domain than in the spatial domain.

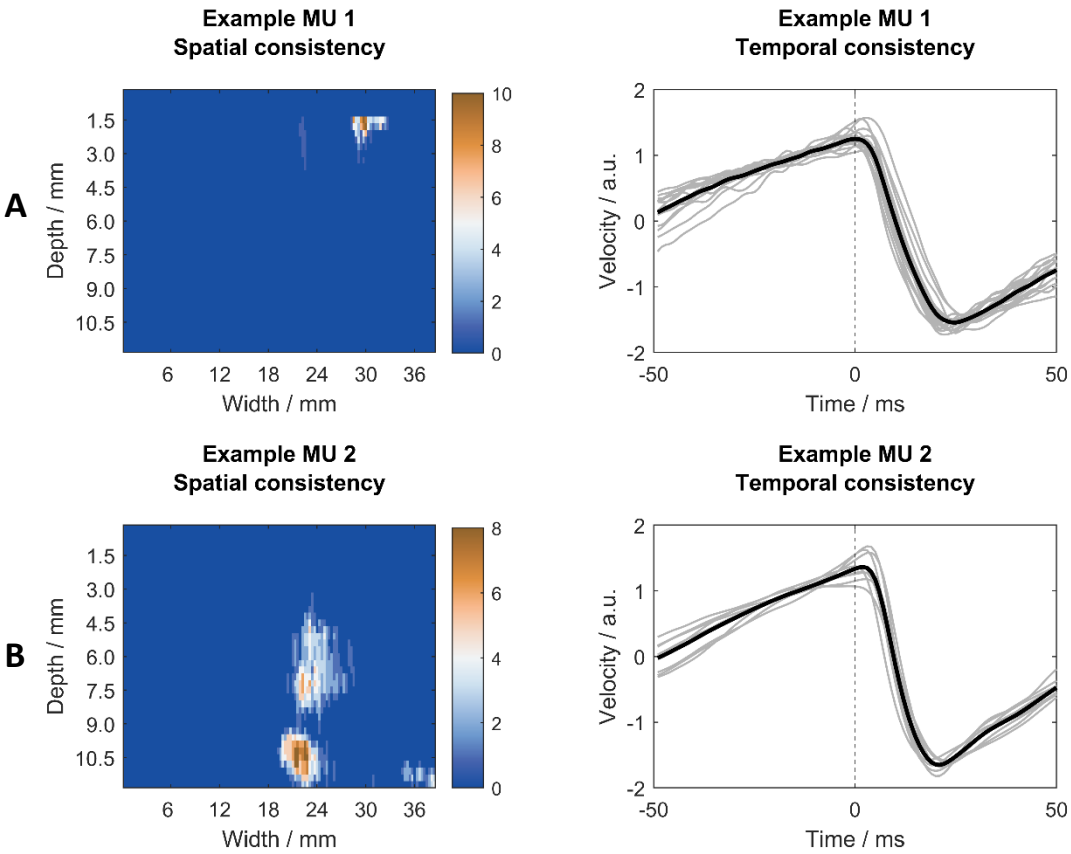

Figure 8. Repeatability across trials for MU identification. Left: motor unit activity maps (MUAMs) for two representative motor units (MUs) across 17 trials (panel A) and 8 trials (panel B) showing high consistency in location of MU motion domain, with some shift as expected due to differing activity level. Each plot is made by overlaying the MU motion domains for all trials containing that MU, such that pixels which are not in the motion domains for and trials have a value of zero. It can be seen from these plots that the regions are very consistent across trials, however there is shifting and morphing. Right: normalised motor unit spike-triggered average (MU-STA) curves for the same example MUs across all trials (grey lines) and average across these trials (black lines). These examples show the high repeatability in space and time.

Mechanical Twitch Duration, Contracting Time and Activation Delay

Fig. 9a shows the histogram of activation delays, with average value 3.8 ± 3.0 ms. A small proportion (3.8% of 390) of calculated activation delays were smaller than zero, suggesting an onset of mechanical motion prior to the electrical AP, due to numerical errors resulting from noise. Fig. 9b shows a histogram of the mechanical twitch duration, with average value 17.9 ± 5.3 ms. Figs. 9c and 9d show histograms for contraction times, with average values 12.1 ± 4.0 ms and 56.6 ± 8.4 ms for the active and total contraction times, respectively. All these measures are shown here for the first time for individual MUs in voluntary contractions.

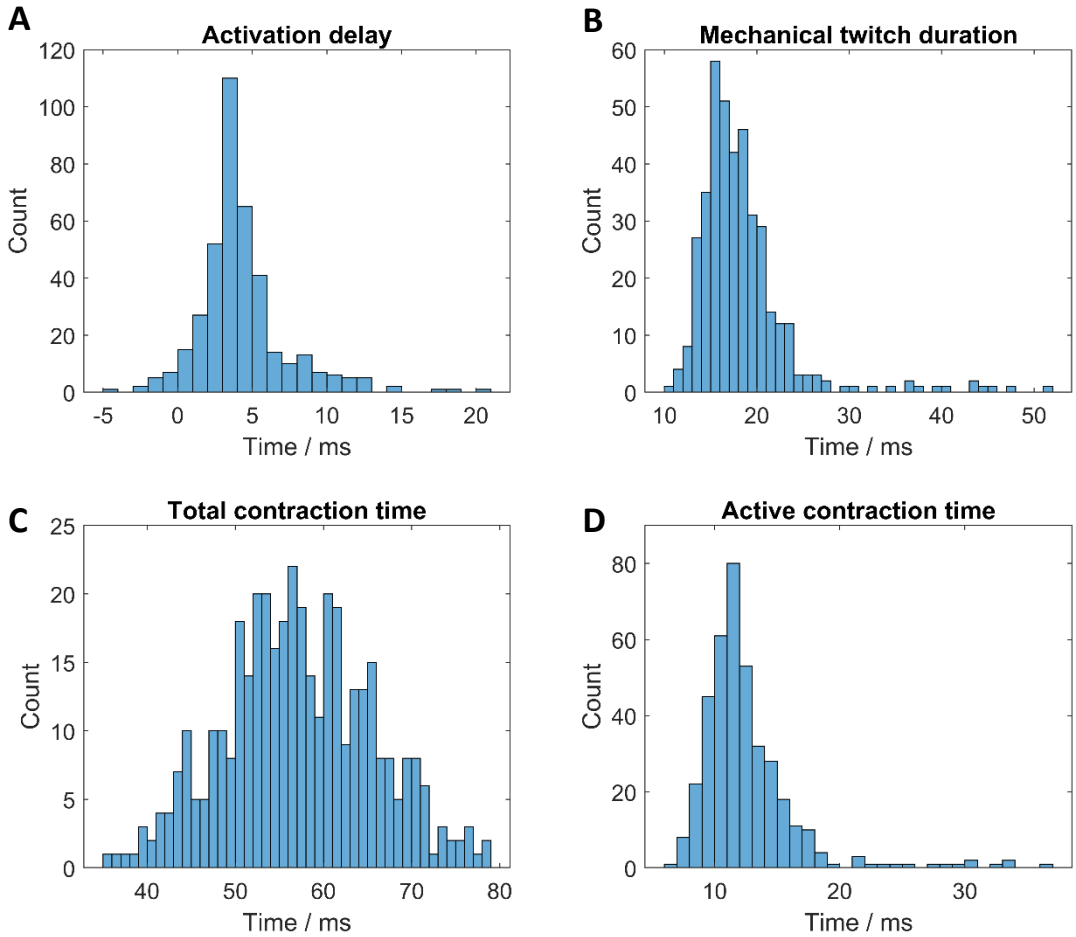

Figure 9. Histograms of temporal features extracted from the velocity profiles of motor unit twitches. A) Histogram of activation delay. B) Histogram of duration of mechanical twitch. C) Histogram of total contraction times. D) Histogram of active contraction times. For all histograms, *n* = 390 from 51 MUs across 10 participants.

Muscle Twisting

In some instances, we observed regions of both positive and negative motion in response to AP discharges. These can be explained by a twisting of the muscle unit, where one group of muscle fibres moves towards the probe next to a group of fibres moving away from the probe. This is the first observation of this phenomenon in natural contractions, and is in agreement with the twisting seen in electro-stimulated contractions (Deffieux *et al.*, 2008). An example is shown in Fig. 10b.

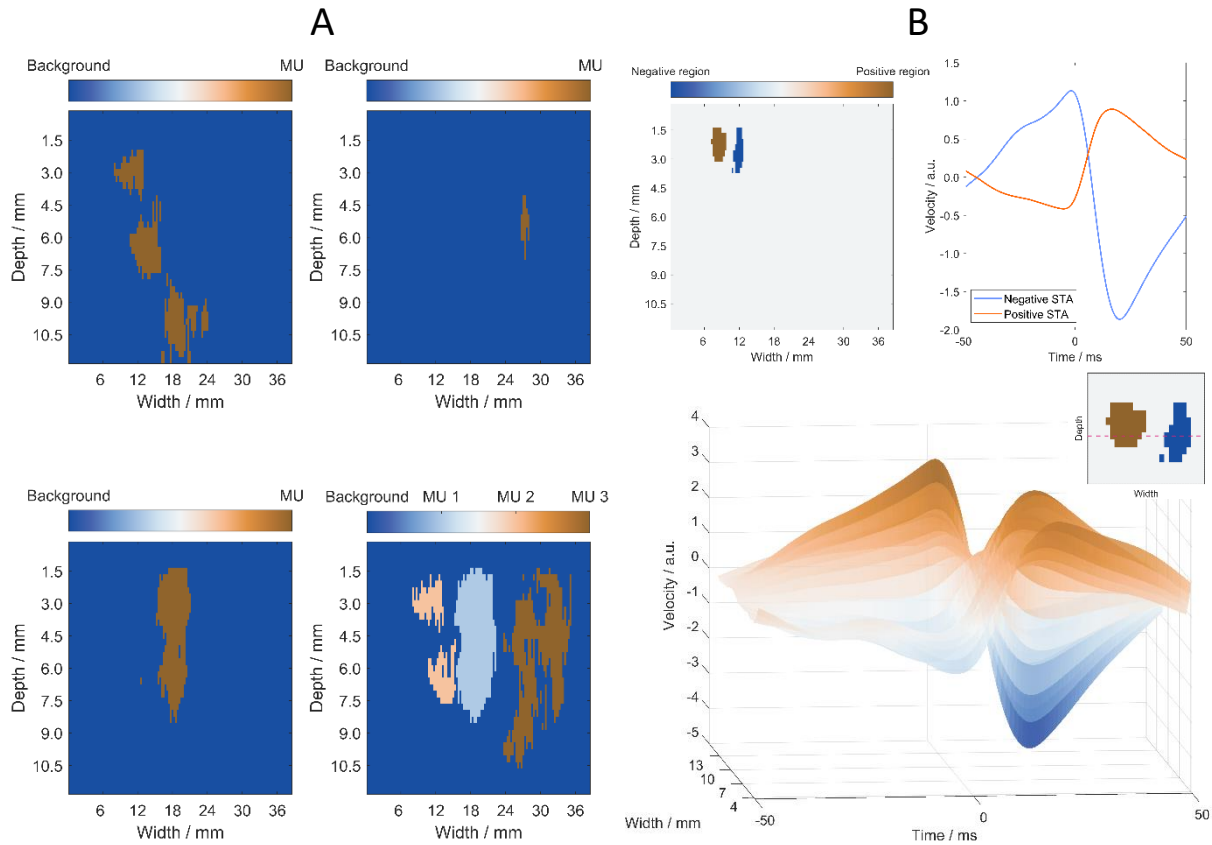

Figure 10. Examples of interesting motor unit (MU) behaviour. A) Different types of MU motion domain identified – these may be split, and vary hugely in size and shape. Also shown is a plot of the motion domains of 3 MUs active in the same trial (which don't overlap in this case). B) An example of 'twisting' – part of the MU motion domain moves up and part moves down in response to a discharge. This is the first time this has been seen in natural neural contractions. Top: MUAM and MU-STAs for the positive and negative moving parts of the motion. Bottom: Mesh plot showing the motion along a line at a depth of 3 mm.

Discussion

We proposed a new technique that allows for the parallel study of the electrical activity and mechanical behaviour of individual MU twitches during voluntary contractions. This method provides new perspectives in physiology for the study of muscular contractions at a granular level in relation to neuromuscular control. Measures presented in this work, including E-C coupling time, show the potentials of the technique. Furthermore, the approach could open up new avenues for the study and monitoring of pathologies that relate to muscle mechanics, and those which relate to the translation of motion from the electrical to the mechanical domains. Furthermore, we have further developed the commonly used STA technique to involve a variability weighting, allowing for identification of regions of not only high motion across

spike instances, but also low variability. This has implications for other applications where relevant signals may be masked, or where long recording times are not possible.

Current methods to study MU motion within a muscle suffer from several limitations. Often, these use electrical stimulation to elicit motion (Deffieux *et al.*, 2008), resulting in contractions different than in natural motor control. Further limitations include the use of systems with low time resolution compared to the durations of the involved physiological processes (Botter *et al.*, 2013), low spatial resolution to properly separate contributions from multiple dispersed MUs (Birkbeck *et al.*, 2020), or the deployment of invasive techniques, impacting the mechanics of the system and enabling the detection of just one MU at a time (Rohlén *et al.*, 2020*a*). In contrast, the technique presented here allows for the non-invasive characterisation of multiple units within a muscle concomitantly in full – from generating electrical impulses, to producing mechanical deformation and movement, thus overcoming the listed limitations in previous work.

Technique validation

The online decomposition algorithm for the HDsEMG has been validated elsewhere, by comparison both with offline decomposition algorithms and with state-of-the-art intramuscular EMG decomposition (Barsakcioglu *et al.*, 2021). As such, by using these decomposition results to process the velocity maps, we ensure that our STA process identifies MU activity from single units. Furthermore, our STA processing involves dividing the velocity maps by the variance across each spike incidence, hence if no consistent movement was present, the processed maps would be uniformly zero. The proposed STA processing ensures that regions we are extracting are specific to the target unit and consistent across discharges; the normalization by the variance in the STA process validates the presence of consistent movements associated to individual target units.

Further, we validated the technique as a whole by analysing the repeatability of the MU-STA twitch profiles, and by comparing the MU motion domains across trials. The low standard deviations of the MU-STA curves found for the MUs across different trials showed the repeatability of the method, despite the *in vivo* nature of the work, and the variation in force level across different trials.

Spatially, MU motion domains were found to have higher similarity to the motion domains of the same MU in different trials than to that of different, simultaneously active, MUs (on average, 47% vs 7% respectively). The former shows high spatial consistency of MU motion

domain location serving as evidence that we are correctly identifying consistent MU motion domains. As expected, this metrics of consistency resulted lower than 100% due to factors such as probe and muscle shifting, translation and muscle morphing due to other active MUs, concomitant activation of different MUs, noise, and thresholding effects.

Although these values show that the method successfully identifies and separates contributions to muscular motion from multiple simultaneous MUs, they also highlight that even constant isometric contractions imply complex dynamics at the muscle unit level resulting from mechanical coupling with neighbouring units. Further evidence for the validity of the technique is the agreement of some of the measured physiological parameters with the corresponding parameters identified with different approaches in previous studies, as discussed in the following section.

Physiology of the muscle unit

Our analysis in the temporal motion domain yielded three mechanical twitch defining parameters: twitch duration, active contraction time, and total contraction time. In their work, Rohlén et al. identified a parameter called 'twitch duration' (approximately 50 ms) (Rohlén *et al.*, 2020*b*) and Deffieux et al. identified a parameter called 'contraction time' (70.1 $\pm$ 2.0 ms) (Deffieux *et al.*, 2008), analogous to our 'total contraction time' (57.2 $\pm$ 8.7 ms). Our results are within one standard deviation of those by Rohlén et al. (Rohlén *et al.*, 2020*b*). Differences among studies may be due to the different muscles investigated, and methodological differences. On the other hand, larger differences in comparison to the results by Deffieux et al. (Deffieux *et al.*, 2008) are likely due to their use of electrical stimulation.

From the time-domain analysis, we extracted the activation delay, i.e. the time between the electrical AP onset and the onset of motion caused by the twitch. We believe that the main contribution to this value is the E-C coupling, and this work therefore represents the first attempt to measure the duration of the E-C coupling in individual MUs in sustained voluntary contractions, which has potential impact for further study of neuropathies, myopathies and other conditions known to affect certain aspects of AP transmission (Orizio *et al.*, 1997; Granata *et al.*, 2000; Kaneko *et al.*, 2002). In electrically stimulated contractions, the time between the stimulation and the onset of fascicle motion is attributed to the combination of E-C coupling and synaptic transmission (6.05 $\pm$ 0.64 ms, (Nordez *et al.*, 2009)), 5.57 $\pm$ 1.37 ms (Hug *et al.*, 2011*a*)). Whilst these studies measure the time between electrical stimulation and fascicle motion, our study instead measures the time between onset of electrical activity as

detected by HDsEMG and onset of fascicle motion, thereby not including the time for synaptic transmission. We would therefore expect smaller latencies in our study as the proposed method accounts for E-C coupling only. Another stimulation-based study instead measured the time between EMG onset and fascicle motion (2.2 ± 0.3 ms (Esposito *et al.*, 2011)), which therefore only accounts for E-C coupling, and is closer to our value of 3.8 ± 3.0 ms.

It is known that the EMD is more than two times greater for voluntary contractions as opposed to stimulated contractions, attributed to the reversed recruitment size principle resulting in faster force development for stimulated contractions (Hopkins *et al.*, 2007). However, some studies suggest that E-C coupling is also greater in voluntary contractions (19.21 ± 6.79 ms (Begovic *et al.*, 2014)), in contrast to our results which well align with those reported for stimulation studies. This may be explained by the use of less precise bipolar EMG, introducing error due to placement at an unknown distance from the motor point, which would be around 10 ms for electrodes placed at 4 cm from the motor point, given that the conduction velocity of fibres is 4 m/s (Hug *et al.*, 2011*b*). Furthermore, the use of A-mode ultrasound to measure the muscle thickness and estimate the onset of fascicle motion may introduce error due to muscle morphing and A-mode placement. This was circumvented in the current study by using B-mode ultrasound with a processing pipeline to identify individual MU motions. Our results suggest that E-C coupling is similar in stimulated and voluntary contractions, and as such the difference in EMD is likely to be attributed to the difference in recruitment order.

Spatially, we defined a 'MU motion domain' as the region of muscle which moves with magnitude and synchronicity above a defined threshold when an AP travels along the muscle unit. Anatomically, MU territories have been defined as the cross-sectional area of the innervated fibres by a MU. MU territories have been studied using glycogen depletion experiments (Edström & Kugelberg, 1968; Garnett *et al.*, 1979; Bodine *et al.*, 1988) and scanning EMG (Stålberg & Dioszeghy, 1991). The latter method analyses the MU territory in one dimension, thus the measured size depends on the location of the needle intersection with respect to the MU cross-section (Gootzen, 1990). Furthermore, 'silent zones' (regions of no EMG signal) are seen in these measurements, which may be a result of the innervation being scattered in different portions of the muscle (Stålberg & Dioszeghy, 1991). More recently, MRI was paired with electrostimulation to measure human MU territories in-vivo (Birkbeck *et al.*, 2020). These studies found MU regions which they categorised as either circular,

elliptical, crescent shape, spider, or split, corresponding well with our MU motion domain shape, which varied largely between MUs (Fig. 10a). The MRI experiments did not detect overlapping MUs due to methodology constraints, whereas we identified overlapping MU motion domains in every participant.

Limitations and Future Applications

While the work presented marks a novel approach to the study of MUs, there are some limitations of note. The results presented only show the technique working on low-force contractions (maximum 10% MVC). Translating the results to higher MVCs is not straightforward as they generate high global motion that obscures the individual MU discharges. Higher forces also increase the likelihood of probe movement and slippage. So far, the processing assumes that the probe remains stationary throughout the recording, which is a valid assumption for low forces; however for higher forces (or dynamic contractions) this would not hold and considerations for its movement would have to be made.

Due to the use of HDsEMG to identify MUAPs, our methodology currently can only detect MUs near the surface of the skin. However, had we used iEMG we would detect a smaller number of individual units within close range of the needle tip (Lowery *et al.*, 2006), and the needle (for the case of needle recordings) would cause mechanical coupling disrupting the results. In principle this can be solved by directly decomposing the US in MU discharges (Ali *et al.*, 2020; Rohlén *et al.*, 2020*a*, 2020*b*), however further validation on the basic assumptions underlying the mechanical components and coupling of MU twitches is necessary.

On the processing side, the autocorrelation velocity tracking algorithm employed is limited to one dimension, however the motion generated by the fibres contractions is multidimensional. Expanding this into a two-dimensional tracking so both forms of the fibre cross-sectional expansion movement can be tracked could be considered. On the same note, the resolution of our US image is limited so individual muscle fibres cannot be detected.

Finally, due to the nature of the US imaging and the materials from which the HDsEMG grids are made, it is not possible to position one on top of the other. While this is not an impediment to the method as the fibres lie longitudinally over the muscle belly, muscle fibres do not span the entire muscle length, so some MUs detected by EMG terminate before reaching the US imaging plane. The use of a HDsEMG array which is invisible to US would enable the probe to be positioned directly above the array, solving the out-of-plane issue.

Some design options for this type of arrays have been proposed (Botter *et al.*, 2013), however they have only been used for gross muscle movements and it is unclear whether they are appropriate for individual MU investigations, as lower SNR and small artifacts may impact the processing of the small-scale movements.

References

Ali H, Umander J & Rohlén R (2020). A Deep Learning Pipeline for Identification of Motor Units in Musculoskeletal Ultrasound. *IEEE Access* **8,** 170595–170608.

Barsakcioglu DY, Bracklein M, Holobar A & Farina D (2021). Control of Spinal Motoneurons by Feedback from a Non-Invasive Real-Time Interface. *IEEE Trans Biomed Eng* **68,** 926–935.

Begovic H, Zhou GQ, Li T, Wang Y & Zheng YP (2014). Detection of the electromechanical delay and its components during voluntary isometric contraction of the quadriceps femoris muscle. *Front Physiol* **5,** 1–8.

Birkbeck MG, Heskamp L, Schofield IS, Blamire AM & Whittaker RG (2020). Non-invasive imaging of single human motor units. *Clin Neurophysiol* **131,** 1399–1406.

Bodine SC, Garfinkel A, Roy RR & Edgerton VR (1988). Spatial distribution of motor unit fibers in the cat soleus and tibialis anterior muscles: Local interactions. *J Neurosci* **8,** 2142–2152.

Botter A, Vieira TMM, Loram ID, Merletti R & Hodson-Tole EF (2013). A novel system of electrodes transparent to ultrasound for simultaneous detection of myoelectric activity and B-mode ultrasound images of skeletal muscles. *J Appl Physiol* **115,** 1203–1214.

Calderón JC, Bolaños P & Caputo C (2014). The excitation-contraction coupling mechanism in skeletal muscle. *Biophys Rev* **6,** 133–160.

Cavanagh PR & Komi P V. (1979). Electromechanical delay in human skeletal muscle under concentric and eccentric contractions. *Eur J Appl Physiol Occup Physiol* **42,** 159–163.

Deffieux T, Gennisson JL, Tanter M & Fink M (2008). Assessment of the mechanical properties of the musculoskeletal system using 2-D and 3-D very high frame rate ultrasound. *IEEE Trans Ultrason Ferroelectr Freq Control* **55,** 2177–2190.

Dieterich A V., Botter A, Vieira TM, Peolsson A, Petzke F, Davey P & Falla D (2017). Spatial variation and inconsistency between estimates of onset of muscle activation from EMG and ultrasound. *Sci Rep* **7,** 1–11.

Edström L & Kugelberg E (1968). Histochemical composition, distribution of fibres and fatiguability of single motor units. Anterior tibial muscle of the rat. *J Neurol Neurosurg*

Psychiatry **31,** 424–433.

Esposito F, Cè E, Rampichini S, Limonta E, Venturelli M, Monti E, Bet L, Fossati B & Meola G (2016). Electromechanical delay components during skeletal muscle contraction and relaxation in patients with myotonic dystrophy type 1. *Neuromuscul Disord* **26,** 60–72.

Esposito F, Limonta E & Cè E (2011). Passive stretching effects on electromechanical delay and time course of recovery in human skeletal muscle: New insights from an electromyographic and mechanomyographic combined approach. *Eur J Appl Physiol* **111,** 485–495.

Garnett RA, O'Donovan MJ, Stephens JA & Taylor A (1979). Motor unit organization of human medial gastrocnemius. *J Physiol* **287,** 33–43.

Gootzen THJM (1990). Electrophysiological Investigation of Motor Unit Structure by means of Scanning EMG. In *Muscle Fibre and Motor Unit Action Potentials*, pp. 89–106.

Granata KP, Ikeda AJ & Abel MF (2000). Electromechanical delay and reflex response in spastic cerebral palsy. *Arch Phys Med Rehabil* **81,** 888–894.

Hopkins JT, Feland JB & Hunter I (2007). A comparison of voluntary and involuntary measures of electromechanical delay. *Int J Neurosci* **117,** 597–604.

Hug F, Gallot T, Catheline S & Nordez A (2011*a*). Electromechanical delay in biceps brachii assessed by ultrafast ultrasonography. *Muscle and Nerve* **43,** 441–443.

Hug F, Lacourpaille L & Nordez A (2011*b*). Electromechanical delay measured during a voluntary contraction should be interpreted with caution. *Muscle and Nerve* **44,** 838.

Ibáñez J, Del Vecchio A, Rothwell JC, Baker SN & Farina D (2021). Only the fastest corticospinal fibers contribute to B corticomuscular coherence. *J Neurosci* **41,** 4867–4879.

Kaneko F, Kiyoshi Onari M, Kawaguchi K, Tsukisaka K & Serge H. Roy (2002). Electromechanical Delay After ACL Reconstruction: An Innovative Method for Investigating Central and Peripheral Contributions. *J Orthop Sport Phys Ther*.

Loupas T, Gill RW & Powers JT (1995). An Axial Velocity Estimator for Ultrasound Blood Flow Imaging, Based on a Full Evaluation of the Doppler Equation by Means of a Two-

Dimensional Autocorrelation Approach. *IEEE Trans Ultrason Ferroelectr Freq Control* **42,** 672–688.

Lowery MM, Weir RFF & Kuiken TA (2006). Simulation of intramuscular EMG signals detected using implantable myoelectric sensors (IMES). *IEEE Trans Biomed Eng* **53,** 1926–1933.

Merletti R & Muceli S (2019). Tutorial. Surface EMG detection in space and time: Best practices. *J Electromyogr Kinesiol* **49,** 102363.

Morimoto S & Takemori S (2007). Initial mechanomyographical signals from twitching fibres of human skeletal muscle. *Acta Physiol* **191,** 319–327.

Nordez A, Gallot T, Catheline S, Guével A, Cornu C & Hug F (2009). Electromechanical delay revisited using very high frame rate ultrasound. *J Appl Physiol* **106,** 1970–1975.

Orizio C, Esposito F, Paganotti I, Marino L, Rossi B & Veicsteinas A (1997). Electrically-elicited surface mechanomyogram in myotonic dystrophy. *Ital J Neurol Sci* **18,** 185–190.

Rohlén R, Stålberg E & Grönlund C (2020*a*). Identification of single motor units in skeletal muscle under low force isometric voluntary contractions using ultrafast ultrasound. *Sci Rep* **10,** 1–11.

Rohlén R, Stålberg E, Stoverud KH, Yu J & Grönlund C (2020*b*). A Method for Identification of Mechanical Response of Motor Units in Skeletal Muscle Voluntary Contractions Using Ultrafast Ultrasound Imaging - Simulations and Experimental Tests. *IEEE Access* **8,** 50299–50311.

Sopher RS, Amis AA, Davies DC & Jeffers JRT (2017). The influence of muscle pennation angle and cross-sectional area on contact forces in the ankle joint. *J Strain Anal Eng Des* **52,** 12–23.

Stålberg E & Dioszeghy P (1991). Scanning EMG in normal muscle and in neuromuscular disorders. *Electroencephalogr Clin Neurophysiol Evoked Potentials* **81,** 403–416.

Tweedell AJ, Tenan MS & Haynes CA (2019). Differences in muscle contraction onset as determined by ultrasound and electromyography. *Muscle and Nerve* **59,** 494–500.

Vasseljen O, Dahl HH, Mork PJ & Torp HG (2006). Muscle activity onset in the lumbar

multifidus muscle recorded simultaneously by ultrasound imaging and intramuscular electromyography. *Clin Biomech* **21,** 905–913.

Del Vecchio A, Holobar A, Falla D, Felici F, Enoka RM & Farina D (2020). Tutorial: Analysis of motor unit discharge characteristics from high-density surface EMG signals. *J Electromyogr Kinesiol* **53,** 102426.

Waasdorp R, Mugge W, Vos HJ, De Groot JH, Verweij MD, De Jong N, Schouten AC & Daeichin V (2021). Combining ultrafast ultrasound and high-density EMG to assess local electromechanical muscle dynamics: a feasibility study. *IEEE Access*1–1.

Waasdorp R, Mugge W, Vos HJ, Groot JHD, De Jong N, Verweij MD, Schouten AC & Daeichin V (2019). Tracking electromechanical muscle dynamics using ultrafast ultrasound and high-density EMG. *IEEE Int Ultrason Symp IUS*2137–2140.

Additional Information

Data Availability Statement

As the datasets acquired are large, the data was not registered in a public database but may be made available, as appropriate and reasonable, upon request.

Competing Interests

None to declare.

Author Contributions

All experiments were performed at the Department of Bioengineering at Imperial College London. E.L.: developed concept, wrote and developed code for control of Verasonics system, for synchronisation, and for post-processing and data analysis, designed custom probe holder, did in vivo experiments, analysed data, wrote first draft of the manuscript. B.S.: developed concept, did in vivo experiments, analysed data, wrote first draft of the manuscript. D.B.: developed concept, developed code for online EMG decomposition. D.F., M.T., J.I.P: developed concept. All authors interpreted the results, revised the manuscript and provided intellectual feedback. All authors have read and approved the final version of the manuscript and agree to be accountable for all aspects of the work in ensuring that questions related to accuracy or integrity of any part of the work are appropriately investigated and resolved. All persons designated as authors qualify for authorship, and all those who qualify for authorship are listed.

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Key Words

electromyogram; ultrasound; ultrafast ultrasound; motor unit; skeletal muscle; excitation-contraction coupling.

Supporting Information

Additional supporting information can be found in the online Supporting Information section. The available files are:

Supplementary Video 1. Spike-triggered average muscle motion due to an individual motor unit discharge.