

MOVEMENTS ON VOICING

Postural and muscular effects of different upper-limb movements on voicing: Pre-registration of confirmatory analyses (**version 3.0, timestamp 30-5-2023**)

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Author note

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Abstract

Rapid upper limb movements can physically impact voicing. This is likely due to biomechanical impulses on chest-wall kinematics which affect subglottal pressures that drive voicing. Such biomechanical interactions can be an important reason why people move their hands in a pulse-like way during speaking, known as gesturing. However, direct measurements of this physical linkage are lacking, preventing an understanding of the vocal interactions with respiratory related muscle activity in response to different limb movement. In this pre-registration, we outline a systematic approach to unravel said biomechanical interactions between gesturing and voicing by measuring activity of key (respiratory-related) muscle units and ground reaction forces. Participants were asked to perform a steady-state vocalization while producing different upper limb movements. We present preliminary evidence, from 2 individuals, that posture-related muscles are important drivers for gesture-vocal interactions, which are more engaged for some movements than others. We also show that some gesture-related muscle activations are associated with positive increases in vocal amplitude, while other muscle synergies may also associate with decreases in the vocal amplitude. Our results provide preliminary evidence for a *dynamic* biomechanical interaction between upper limb movement and the voice - different gestures affect the voice in different ways depending on the muscle synergy recruited to perform the movement. We pre-register confirmatory analyses, supported by a simulation-based power analyses, for a full study that has been performed with 17 participants.

Keywords: voice, gesture, upper limb movement, postural control, biomechanics

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Introduction

In principle, any muscle that attaches to the rib cage can affect rib cage kinematics and thus affect respiration. Consequently, there are many *potential* respiratory muscles, including those around the upper trunk (e.g., pectoralis major), abdomen (rectus abdominus), and back (erector spinae, seratus posterior/anterior). However, in breathing and speaking without much additional body movements, only a subset of the possible muscles is used. Most notably, the diaphragm and the muscles between the ribs (intercostalis) drive passive speech-supporting respiration (Levangie and Norkin, 2011; Seikel, Drumright, & Hudock, 2019). Only on rarer occasions, when coughing, shouting, or breathing deeply, humans recruit other so-called “accessory” respiratory muscles such as the abs and pectoral muscles (Aliverti, 2016; e.g., Lasserson et al., 2006; for an overview see Seikel et al., 2019). From this it is tempting to conclude that there is only a small set of ‘primary’ respiratory muscles.

Yet, when humans speak or sing, they often move their upper limbs expressively at the same time — called gesturing (Pearson & Pouw, 2021; Wagner, Malisz, & Kopp, 2014). Such upper limb movements recruit a whole range of upper body muscles, including those involved in maintaining posture (e.g., Cordo & Nashner, 1982). Several of these muscles attach to the rib cage (e.g., ab- and pectoral muscles). These muscles are classically listed as accessory to respiratory functioning (Seikel et al., 2019). Therefore, on a second glance, the ‘accessory’ muscles, supposed to only activate during boundary conditions of breathing and speaking, are actually *often* recruited for the sake of gesturing in everyday-ways of speaking. Gestures thus have potential to interact with vocalization, but generally only considered to serve visual communication.

Considering the bare mechanics, even when a gesture is intended for visual communication rather than to affect respiratory drive, if it recruits respiratory-related muscles it will still affect respiratory drive because small changes in rib cage movements will affect subglottal pressure that drive vocal fold vibrations. Given that speaking requires subtle

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modulations of subglottal pressure (Rubin, LeCover, & Vennard, 1967; Johan Sundberg, Elliot, Gramming, & Nord, 1993; Sundberg, Titze, & Scherer, 1993), speaking must be coordinated with the gesture (for an overview see Pouw & Fuchs, 2022).

In this study we will directly assess how upper limb movements are integrated with vocal actions. We do this by monitoring how different, posture-stabilizing, muscle activations for limb movement are related to respiratory-vocal functioning.

The respiratory system in action

The lungs do not move themselves. They passively follow movements of the surrounding structures: the chest wall and the diaphragm. The lungs follow these movement due to negative pressures in the fluid-filled “pleural” space between the lungs and the surrounding structures (Seikel et al., 2019). The muscles that move the surrounding structures of the lungs, and indirectly the lungs themselves, are identified as the primary respiratory muscles. This includes the diaphragm muscle, which by contracting flattens its otherwise dome-like surface, thereby increasing lung volume and leading to inspiration. Further, the intercostal muscles move the chest wall with inspiratory (external intercostalis) and expiratory effects (inner and internal intercostals). When the chest-wall is fully inflated (or deflated), there are passive recoil forces of the ribs (and connective tissues) that will pull the rib cage back towards its equilibrium point. This also means the chest wall can (and often does) move without a primary drive from muscle activations during early phases of expiration and inspiration. Yet, even though elastic recoil forces can be the “prime mover” that produce an expiratory drive, more fine-grained inspiratory-expiratory modulations are simultaneously performed by a complex patterning of muscle activity that dynamically controls the voice (e.g., McNulty, Gevirtz, Hubbard, & Berkoff, 1994).

There are a range of muscles that can be associated with both gesture and respiration. These muscles can be divided into muscles that are involved in making anticipatory postural adjustments and into the prime movers of the arms and muscles. Most well-known postural muscles that may also affect respiration, are the abs (including, rectus abdominus, transversus abdominus). They attach to the rib cage and depress the rib cage when tensioned, thereby increasing intra-abdominal pressure and elevating the diaphragm. But also more ‘focal’ muscles,

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that support upper limb movement, like those in the shoulder girdle region, have been associated with higher expiratory flow rate during coughing (Lasserson et al., 2006) as well as cough sound intensity (McGuinness, Ward, Reilly, Morris, & Smith, 2018). If these muscles activate due to upper limb movements, we should expect constraining effects on vocalization. We emphasize constraint as muscles do not function independently, but in an orchestrated manner. This means that the same increase in say the pectoralis muscle activity may be associated with different respiratory effects.

Interestingly, in non-human animals, pectoral limb related muscle activity has been found to integrate with respiratory action (Cooper & Goller, 2004; Lancaster, Henson, & Keating, 1995). For example, in flying echo-locating bats (*Pteronotus parnellii*) muscles activated for flying (pectoralis) are at the same time mechanically driving expiration for echo-vocalizing. Though during non-flying vocalization, other muscle units will drive expiratory flow (Lancaster et al., 1995). Integration by counteraction has also been observed: When Brown-headed Cowbirds (*Molothrus ater*) perform their wing-displays these are coordinated with their vocalization, but when wing-displays become too vigorous, these birds have to cease vocalization all-together (Cooper & Goller, 2004). In sum, from non-human animal communication research we know that there is an integration of vocalization with respiratory-constraining muscles, even when said muscles are activated for the sake of something else, such as locomotion or for impressing a potential mate (Pouw & Fuchs, 2022). We expect that the upper limbs in humans are similarly integrated with vocalizations.

The respiratory system during upper limb signifying action

There is considerable evidence that human gesture movements with a beat-like or pulsing quality are temporally integrated with vocal aspects of speech (Wagner et al., 2014). For example it is often reported that effortful moments of gesture are strategically timed with excursions in the fundamental frequency (F0) and intensity of speech which are perceived as so-called accented moments of speech (Wagner et al., 2014). This synchronization of pulsing gestural movements with prosodically-relevant vocal inflections in speech (e.g., see [here](#)) is referred to as multimodal prosody (Ginosar et al., 2019; Wagner et al., 2014). Multimodal prosody is traditionally studied as a purely linguistic phenomenon (McClave, 1998). It has recently been argued that this

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phenomenon might be a biomechanical integration: gestural movements might be intended to affect vocal action via respiratory alignment (Pouw, de Jonge-Hoekstra, Harrison, Paxton, & Dixon, 2020; Pouw, Harrison, & Dixon, 2020; Pouw, Harrison, Esteve-Gibert, & Dixon, 2020; Pouw, Paxton, Harrison, & Dixon, 2020).

According to this gesture biomechanics thesis, synchronization is physically driven by sudden acceleration or deceleration of an upper limb segment which recruits focal and/or peripheral muscles that interact with respiratory-vocal control (Pouw & Fuchs, 2022). An exemplar *focal* muscle (or prime mover) would be the pectoralis major that drives internal rotation of the upper arm (humerus). The activation of this chest muscle is in other occasions associated with (forced) inspiratory action (Pettersen, 2006). Furthermore, upper limb actions are often coordinated with a whole suit of muscle units that anticipate and correct the destabilizing effects of limb movement on postural balance ((Baldissera & Tesio, 2017; Cordo and Nashner, 1982; Bouisset and Do, 2008; Baldissera, Rota, and Esposti, 2008). Notably, posture-stabilizing muscles, like the abs (rectus abdominus) are on other occasions implicated in (forced) expiration (LoMauro & Aliverti, 2019).

Gesture-speech biomechanics research has so far focused on kinematics instead of kinetics research (for a review see Pouw & Fuchs, 2022). More extreme peaks in the acceleration of the upper limb movement relate to larger chest-circumference changes, thereby affecting subglottal pressure, and thereby to more extreme acoustic effects on the intensity of vocal sound. Though sometimes an effect on the fundamental frequency (F0) of concurrent voicing is found, it is often less affected than intensity, which is explainable by the fact that F0 is primarily controlled by the larynx as opposed by expiratory flow rate alone. Furthermore, acoustic effects of upper limb movements are more extreme when subjects are in a more unstable standing compared to sitting position. This ties in the idea that a physical impulse (related to force = mass x acceleration), impacts posture (especially when standing), recruiting respiratory-related muscles (that change chest-circumference), which impacts respiratory-vocal functioning (such that intensity and F0 are affected). These previous studies assessed continuous voicing, mono-syllable utterances, and fluent speech production.

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Current pre-registration

The gesture-speech biomechanics thesis hedges on (posture-stabilizing) muscle activations to affect respiratory-vocal functioning. Such muscle activity has however never been directly measured in this context. In this study we investigate through surface electromyography (sEMG) the muscle activity during different upper limb movements which are produced during vocalization while the participant is standing. We will also assess the posture-stabilizing role of some muscles, by relating their activity to ground-reaction forces.

Participants perform no movements ([see here](#)) as well as 4 actions along the sagittal (flexion [see here](#)] and extension [see here](#)]) and transversal plane (internal [see here](#)] and external rotation [see here](#)]) with the elbow flexed at a 90 degree angle. As we know that gesture-speech biomechanics is modulated by posture (Pouw, Harrison, & Dixon, 2020). Therefore, we measure two postural related muscles: the rectus abdominus and the erector spinae muscle group in the lower back. Both the abs and erector spinae directly attach to the rib cage. We also assess focal muscles with and without well-known respiratory effects: We measure an internal rotator muscle on the chest (pectoralis major), and an external rotator muscle on the scapula (infraspinatus). Only the pectoralis major is associated with respiratory interactions as reviewed above and directly attaches to the rib cage. While the infraspinatus is not directly attaching to the rib cage, it does insert onto the scapula which is a dynamic stabilizer for the entire shoulder girdle region (Levin, 1997). Thus, indirectly it might still have a role in gesture-speech biomechanics, but most likely, weakly so.

The key research question and analyses we pre-register here, is whether different actions have different acoustic effects that can be explained by biomechanical factors such as specific muscle activation and the posture-stabilizing role therein. Specifically, if we hypothesize that there are expiratory- and inspiratory-related modulation effects on vocalizations *which can be attributed to the different muscle activation patterns they elicit*. If this is confirmed, it is shown that moving the arms dynamically changes the voice in a much more complex way (Ravignani & Kotz, 2020) than simply *increasing* expiratory flow.

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We support this pre-registration with a preliminary data analyses and power analyses that motivates the eventual confirmatory analyses on a dataset already collected with 17 participants. However, in the final publication we will also address further exploratory questions that are not described in this pre-registration.

Methods

The method section will have a general subsection under each heading reflecting information applicable to both the pilot and planned pre-registered study, and a subsection for the pilot and planned study separately to reflect any deviations.

Participants (General)

The planned and current study has been approved by the Ethics Committee Social Sciences (ECSS) of the Radboud University (reference nr.: 22N.002642).

Pilot study. For the current pilot experiment supporting the pre-registration, the first author (male; right-handed; age 35; 21.7 BMI; 68 kg; length 173cm; upper arm circumference = 31.5cm; triceps skinfold = 16mm) and a female volunteer (right-handed; age 37; 57 kg; 21.5 BMI; length 161cm; upper arm circumference = 26cm; triceps skinfold = 21mm) performed the experiment.

Confirmatory study. We based the number of participants on a power analysis, also reported below. With 80% to 90% power for the two main confirmatory analyse we found that about 15 participants are needed (see below for detail). We have therefore collected a dataset of 17 participants.

With regards to exclusions, to ensure reliability of the surface EMG measurements we only admitted participants with a BMI lower than 25. Further, we only recruited participants that reported not motor impairments.

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Study design

The exploratory and confirmatory study involves a two-level within subject ~~factor~~ wrist-weight manipulation (no weight vs. weight), a two-level within subject vocalization condition (expire vs. vocalize), and a five-level within-subject movement condition ('no movement', 'extension', 'flexion', 'external rotation', 'internal rotation'). With 4 trial repetitions over the experiment, we yield 80 (2 weights x 2 vocalizations x 5 movements x 4 repetitions) trials per participant. Trials were blocked by weight condition and vocalization condition (so that weights and task did not switch from trial to trial). Within blocks all movement conditions were randomized.

Measurement and equipment

Body measurements. To enable future analyses of possible modulating individual-specific body properties we collect some basic information about body properties. Namely, weight, under arm length, upper arm length, triceps skinfold, and upper arm circumference.

Experiment protocol

The experiment was coded in Python using functions from PsychoPy (Peirce et al., 2019). The experiment was controlled via a Brainvision Button Box (Brain Products GmbH, Munich, Germany), which was also streaming its output to the data collection PC unit.

Wrist weight. To manipulate the mass set in motion, we applied a 1 kg wrist weight (Tunturi sports).

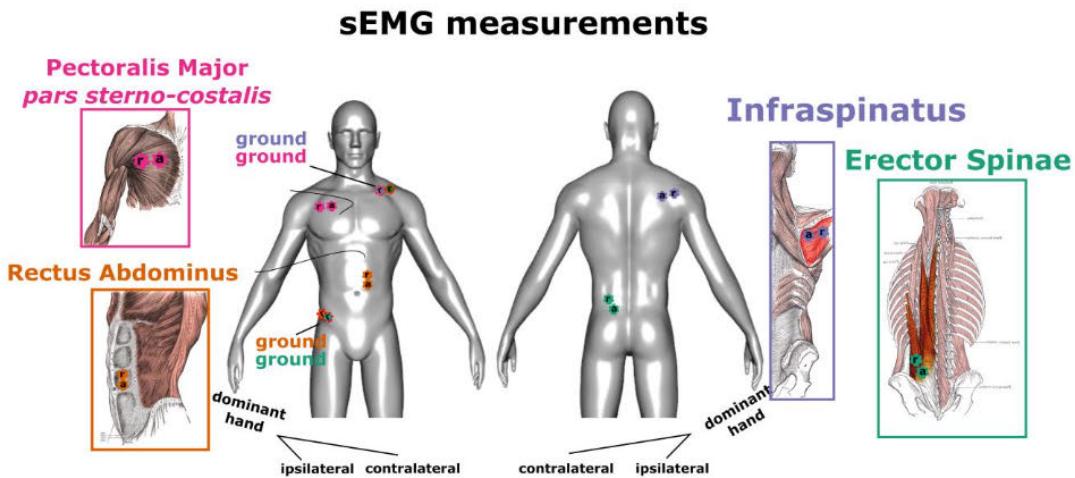
Video and kinematics. The participants are recorded via a videocamera (Logitech StreamCam), sampling at 60 frames per second. We used Mediapipe (Lugaresi et al., 2019) to track the skeleton and facial movements, which is implemented in Masked-piper which we also use for masking the videos (Owoyele, Trujillo, Melo, & Pouw, 2022). The motion-tracked skeleton, specifically the wrist of the dominant hand, is used to estimate movement initiation, peak speed, and the end of the movement. The motion tracking is however only used for determining movement windows and is not central to our analyses.

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Muscle activity (Surface ElectroMyoagrpahy: sEMG). We measured sEMG using a wired BrainAmp ExG system (Brain Products GmbH, Munich, Germany). Disposable surface electrodes (Kendall 24mm Arbo H124SG) were used, and for each of the four target muscles we had 3 (active, reference, ground) electrodes (12 electrodes total). The sEMG system sampled at 2500 Hz (for post-processing filters see below).

For an overview of the electrode attachments see Figure 1. We prepare the skin surface for EMG application with a scrub gel (NuPrep) followed by cotton ball swipe with alcohol (Podior 70%). Active and reference electrodes were attached with a 15mm distance center to center.

Figure 1. Overview sEMG target muscles



Note. Active (a) and reference (r), and ground (g) sEMG electrodes, for each muscle target.

We attached electrodes for focal muscles which directly participate in the internal (pectoralis major) and external rotation (infraspinatus) of the humerus. Electrodes were applied for focal muscles ipsilaterally to the dominant hand. We attached electrodes to the muscle belly of the clavicular head of the pectoralis major, with a ground electrode on the clavicle on the opposite side.

We also attached electrodes for postural muscles which will likely anticipate and react to the postural perturbations due to upper limb movements. Since these muscles should act in the

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opposite direction of the postural perturbation of the dominant hand, we applied electrodes contralaterally to the dominant hand. We attach electrodes to the rectus abdominus, with a ground electrode on the iliac crest on the opposite side. We also attached electrodes to the erector spinae muscle group (specifically, the iliocostalis romborum).

Ground reaction measurements

We used an inhouse-built 1m² balance board with vertical pressure sensors. The sensors were derived and remodified from the Wii-Balance board sensors. The sampling rate was 400Hz. The system was time-locked within millisecond accuracy with the EMG system and the spatial accuracy for locating the center of pressure is below 1mm. A national instruments card, USB-62221 performed the A/D conversion and was connected via USB to the PC.

Acoustics

To ensure proper acoustic intensity measurements we used a headset microphone; MicroMic C520 (AKG, Inc.) headset condenser cardioid microphone sampling at 16Khz. The gain levels of the condenser power source were set by the hardware (and could not be changed).

Recording setup and synchronization. We use LabStreamLayer (<https://github.com/sccn/labstreaminglayer>) which provides a uniform interface for streaming different signals along a network, where a common time stamp for each signal ensures sub-millisecond synchronization. We used a Linux system to record and stream the microphone recordings. Additionally, a second PC collected video, and streamed ground reaction forces, and EMG. A data collection PC collected the audio, ground reaction force, and EMG streams and stored the output in XDF format for efficient storing of multiple time-varying signals.

Pilot study. We post-synchronized the video with the audio recording, using audio-waveform alignment (Pouw, Trujillo, & Dixon, 2020).

Procedure

Participants are admitted to the study based on exclusion criteria (<25BMI and no physical disabilities) and sign an informed consent. We ask participants to take off their shoes

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and we proceed with the body measurements, while informing the participant about the nature of the study. Next, EMG-electrodes were installed, cleansing the skin at the muscle site with abrasive gel and alcohol, and placing the differential electrodes along the muscle fiber direction with an interelectrode distance of 15mm. See Figure 2 for the sEMG electrode locations. The procedures up to the start of the experiment takes about 20 minutes or less in total.

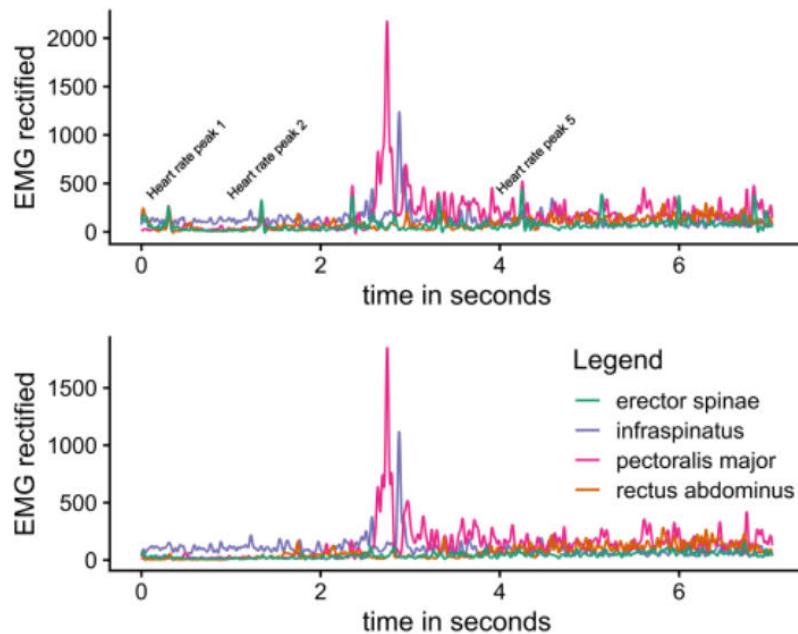
Upon start of the experiment participants take a standing position on the force platform with their feet placed on the two foots printed on the platform. The experiment commences with calibration and practice trials. First 10 seconds of silent breathing without body movements are recorded. Then participants are asked to take a maximum inspiration followed by a maximum expiration to measure signal conditions under respiratory boundary conditions. Then, for the practice trials, every upper limb movement condition was practiced while expiring or during vocalization. The participant is also introduced to wearing the wrist weight of 1kg. After practice trials, participants performed 80 blocked trials.

For each (practice) trial participants are closely guided by the information on the monitor placed at a height of about 75 cm. Firstly, participants are shown a movie clip of the movement to be performed for that trial while they are standing with both arms alongside the body (rest position), and have to prompt the experimenter that they are ready to continue. Then participants are instructed to adopt the start position of the movement, which is a 90-degree elbow flexion, with either an externally rotated humerus (start position for internal rotation), or a non-rotated humerus with the wrist in front of the body (external rotation and elbow extension condition). For the elbow flexion and no-movement conditions the start position equals the rest position. Upon trial start, participants are asked to inhale deeply with a timer counting down from 4 seconds (also shown on the monitor). Then, participants are asked to start ‘vocalizing’ or ‘expiring’. The vocalization is required to be held at a constant intensity, even during movements. After 3 seconds the instruction to perform the movement is given, with visual guidance where the movement end position is so that participants are reminded of the movement. After an additional 4 seconds the trial ends, which allows more than enough time to perform the movement and stabilize vocalization after the perturbation. In the no movement condition, a prompt is given to maintain the rest posture.

Preprocessing of the data streams

EMG. To reduce heart rate ~~artefacts~~, we apply a common method (Drake & Callaghan, 2006) of high-pass filtering the signal at 30Hz using a zero-phase 4th order butterworth filter. We then full-wave rectified the EMG signal and applied a zero-phase low-pass 4th order Butterworth filter at 20Hz. When filtering any signal, we pad the signals to avoid edge effects. We z-normalized the EMG signals within participants before submitting to analyses.

Figure 2. Example of smoothing settings for EMG signals



Change in center of pressure. We upsampled the balanceboard from 400Hz to 2500 Hertz. We then applied a zero-phase low-pass 20Hz 2nd order Butterworth filter to the padded signals. As a key measure for postural perturbation we computed the change center of pressure (L2 norm of the change in the center of pressure in the horizontal and vertical dimension), hereinafter COPc.

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Acoustics. For acoustics we extract the smoothed amplitude envelope (hereonafter envelope) and the Fundamental Frequency (hereonafter F0). For the envelope we apply a Hilbert transform to the waveform signal, then take the complex modulus to create a 1D timeseries, which is then resampled at 2500Hz, and smoothed with a 12Hz Hanning window. We z-normalize the amplitude envelope signals within participants before submitting to analyses.

Though not under consideration here, we will also compute F0 estimates with a window length of 0.005 seconds K. Schaefer-Vincent periodicity detection algorithm. For the F0 estimation we set for males the F0 range to 70-300Hz, and for females to 100-450Hz. We upsample the signal to 2500Hz. We smoothed uninterrupted runs of F0 with a Kolmogorov-Golai (a type of moving average filter) with a span of 111 (44 ms) and order of 4.

Data aggregation. All signals were sampled at, or upsampled to, 2500 Hz. Then we aggregated by aligning time series in a combined by-trial format to increase ease of combined analyses. We linearly interpolated signals when sample times did not align perfectly.

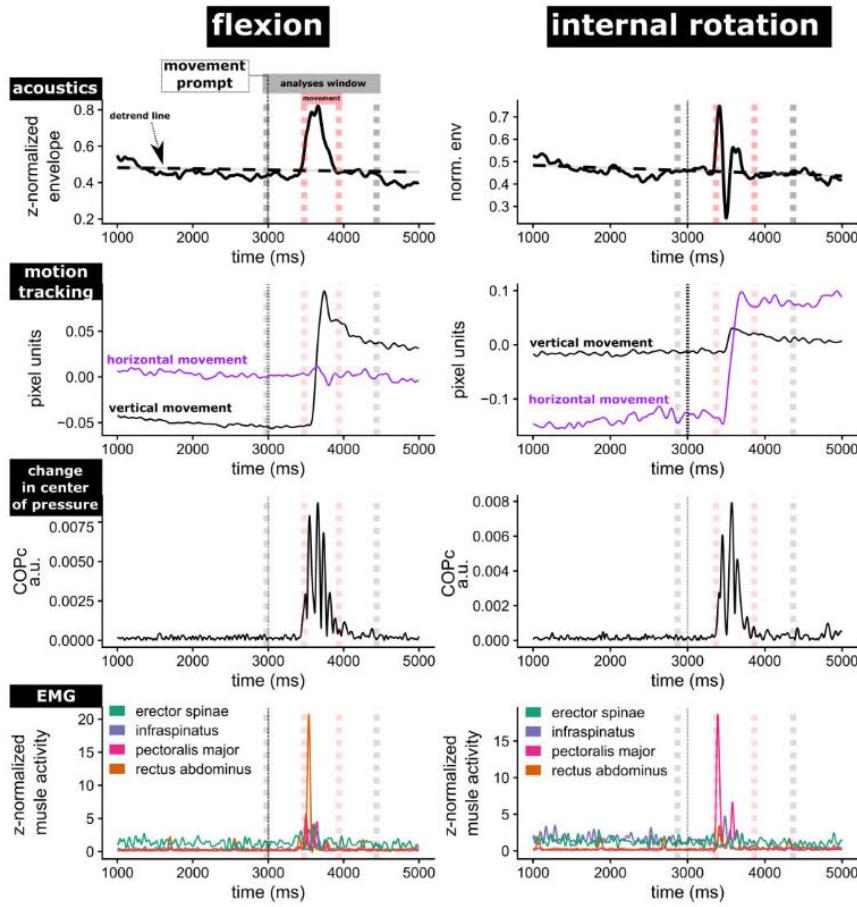
Data sharing & Privacy. Video data is deidentified using the masked-piper tool to mask faces and body while maintaining kinematic information (Owoyele et al., 2022).

Overview data and analyses

Figure 3 provides an example of two trials of a participant vocalizing. Note that there is a general decline in the amplitude of the vocalization during a trial (see Figure 3), due to the subglottal pressure falling when the lungs deflate. To quantify deviations from stable vocalizations, we therefore detrend the amplitude envelope timeseries and assessed positive or negative peaks relative to this trend line. For the acoustic envelope, muscle activity, and the change in center of pressure we will determine the global maxima happening within the analyses window (i.e., within a trial we take a local maximum occurring between movement onset and offset based on the speed time series of the wrist motion tracking data). We will analyze positive and negative peaks in acoustics separately.

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Figure 3. Time series showing acoustic amplitude envelope (z-normalized), motion tracking of the wrist (pixel units), change in center of pressure (arbitrary units), and EMG (z-normalized)



Note. Two example trials and their associated signals are shown, for a flexion movement condition (left panels) and internal rotation movement condition (right panel). At time = 0 (not shown) the prompt is given to the participant to vocalize. We determine a detrending line using linear regression (dashed line) for the 1 to 5 seconds after the vocalization prompt. After 3 seconds (3000ms) of vocalization there is a movement prompt. Based on wrist speed (L2 norm of changes in horizontal and vertical movement) we determine the window where we asses peaks in our signals (the trough before a wrist speed peak, and the trough following the wrist peak speed). In these trials, the analyses window is indicated by grey dashed bars. The red dashed lines indicate the onset and offset of a movement.

Results

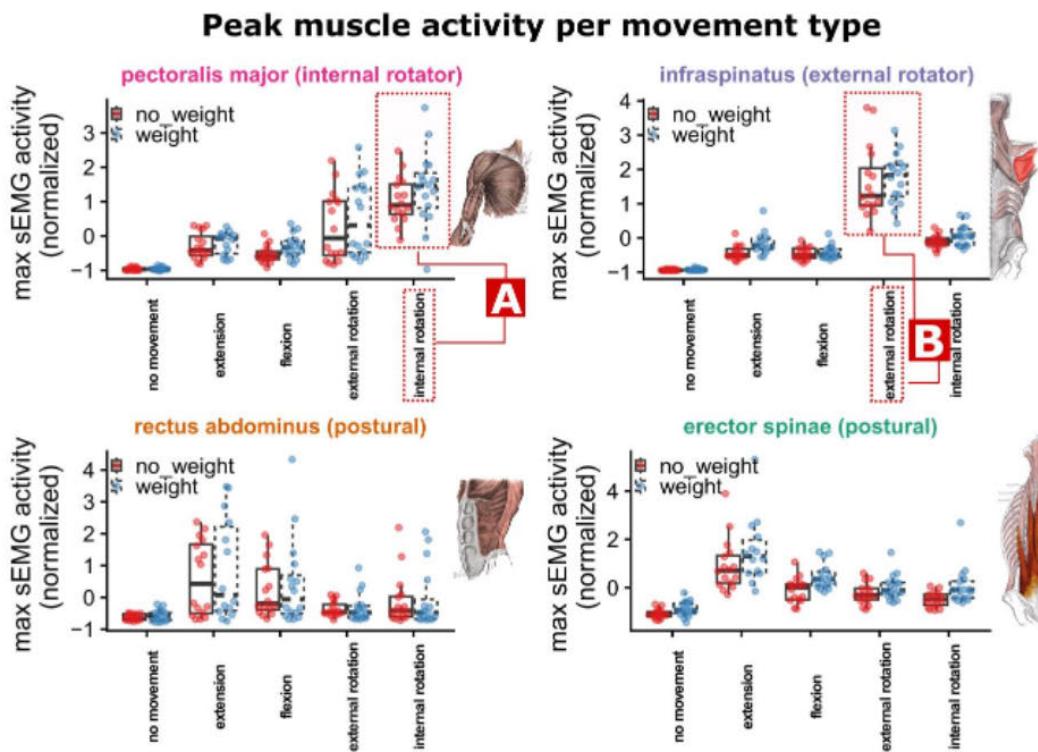
In the following we report preliminary results from the pilot study analyses. In preparation for the key ‘confirmatory’ analyses we also report power analyses based on simulations of the current pilot data (Kumle, Võ, & Draschkow, 2021).

Descriptive results

Figure 4 provides an overview of the peak muscle activity for each movement condition, for the two wrist weight conditions separately. Figure 4 clearly shows that pectoralis and infraspinatus are mainly involved in rotation of the humerus, as an agonist-antagonist pair. In contrast, rectus abdominus and erector spinae, although not directly generating elbow flexion or extension, seem to stabilize the body in the fore-aft direction. Table 1 provides the numerical information of figure 4. Table 2 provides the correlations of the peaks in muscle activity. We can confirm that indeed the focal muscles powering internal (pectoralis major) or external (infraspinatus) rotation are peaking in activity during these movement conditions.

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Figure 4. Peak muscle activity per movement type



Note. A. Pectoralis activity for the 5 movement conditions, separately for the ‘no weight’ and ‘weight’ condition, collapsed for the vocalization and expiration conditions. Dots represent individual trials, for both subjects. B. Infraspinatus activity for the external rotation movement condition.

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Table 1. Normalized peak muscle activity for the different movement conditions

muscle	Movement condition	Mean	lowCI	hiCI	sd
erector spinae	no movement	-0.985	-1.084	-0.886	0.049
erector spinae	extension	1.211	0.748	1.675	0.226
erector spinae	flexion	0.182	-0.031	0.394	0.104
erector spinae	external rotation	-0.119	-0.297	0.058	0.087
erector spinae	internal rotation	-0.212	-0.451	0.026	0.117
infraspinatus	no movement	-0.930	-0.939	-0.921	0.005
infraspinatus	extension	-0.304	-0.419	-0.188	0.056
infraspinatus	flexion	-0.440	-0.508	-0.372	0.033
infraspinatus	external rotation	1.671	1.348	1.994	0.158
infraspinatus	internal rotation	-0.016	-0.111	0.079	0.047
pectoralis major	no movement	-0.958	-0.973	-0.942	0.007
pectoralis major	extension	-0.256	-0.390	-0.121	0.066
pectoralis major	flexion	-0.423	-0.536	-0.309	0.056
pectoralis major	external rotation	0.402	0.023	0.780	0.186
pectoralis major	internal rotation	1.219	0.887	1.550	0.162
rectus abdominus	no movement	-0.600	-0.652	-0.546	0.026
rectus abdominus	extension	0.738	0.224	1.250	0.250
rectus abdominus	flexion	0.328	-0.078	0.733	0.199
rectus abdominus	external rotation	-0.341	-0.470	-0.212	0.063
rectus abdominus	internal rotation	-0.078	-0.390	0.232	0.152

Note. These are the descriptive results associated with Figure 4. The values indicate the peak EMG activity z-normalized for each muscle, and aggregated over participant and trials.

Further, postural muscles such as the rectus abdominus are especially active for extension movements (and secondarily flexion). Confirming their combined postural role, the muscle activity of the rectus abdominus and erector spinae are reliably correlated (see Table 2). Confirming their antagonistic role in rotating the humerus, the pectoralis and infraspinatus muscle activity are also reliably correlated, indicative of their joint agonist/antagonistic control of posture and upper limb rotation, respectively. Lastly, from Table 3 it is clear that adding a wrist weight increases the peak muscle activity for all muscles (with no overlapping 95% confidence intervals in the weight vs. no weight condition).

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Table 2. Correlations between peak muscle activity

	1	2	3
1. pectoralis major	-		
2. infraspinatus	.35***	-	
3. rectus abdominus	.07	-.04	-
4. erector spinea	.12	.08	.36***

Note. Pearson correlations between peaks in muscle activity. * < .05, ** < 0.01, *** < 0.001.

Table 3. Normalized peak muscle activity per weight condition and movement condition

muscle	weight_condition	M	lowCI	hiCI	sd
erector spinea	no weight	-0.202	-0.397	-0.006	0.098
erector spinea	weight	0.206	-0.037	0.449	0.122
infraspinatus	no weight	-0.066	-0.290	0.159	0.113
infraspinatus	weight	0.067	-0.156	0.291	0.112
pectoralis major	no weight	-0.097	-0.296	0.105	0.100
pectoralis major	weight	0.098	-0.148	0.344	0.123
rectus abdominus	no weight	-0.052	-0.241	0.137	0.095
rectus abdominus	weight	0.054	-0.202	0.310	0.128

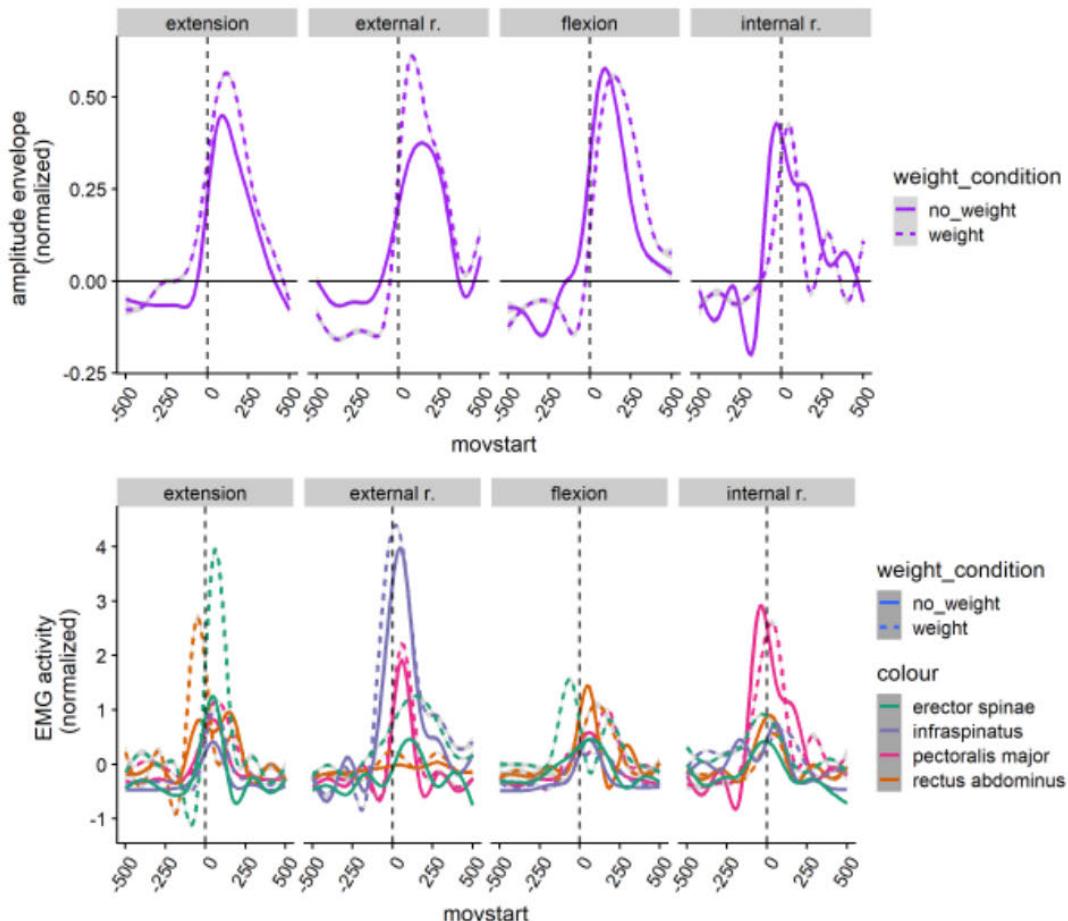
Note. The values indicate the peak EMG activity (normalized per muscle) per weight condition.

Exploratory visualization

Figure 5 shows the acoustic amplitude envelope (in purple, upper panel) and EMG activity over a 1-second interval around movement onset. This plot will serve as the rationale for the confirmatory analyses. Firstly, the acoustic envelope shows a clear positive peak during movement, relative to the trend line for that vocalization, that may or may not be *preceded* by a less extreme negative peak. Especially, for the internal rotation we observe such a negative peak. We will therefore assess whether particular movement conditions predict higher magnitude negative and positive peaks in the amplitude envelope (analysis 1). We further assess whether particular muscles predict the magnitude of positive or negative peaks (analysis 2). Finally, we will assess whether particular muscles are related to postural stability, to confirm the posture-mediating role of the muscles (analysis 3). While this pre-registration and pilot analyses report on the peaks in the signals (in part to support a more straightforward power analyses), in the confirmatory study we plan to perform continuous trajectory analyses using generalized additive modeling similar to previous work (Pearson & Pouw, 2022).

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Figure 5. GAM, a type of smoothed non-linear regression of the acoustic envelope, and muscle activity



Note. GAMs, a type of smoothed non-linear regression over time, where time is centered at 0 at the movement onset (as determined by the motion tracking of the wrist). The smoothed conditional means are generated by fitting non-linear smooths using generalized additive model in R-package ‘ggplot2’. It can be seen that especially for the extension movement there are clear anticipatory postural muscle activations (Aruin & Latash, 1995), of the rectus abdominus before the movement onset, which is then followed by postural adjustments of the erector spinae. For the flexion condition, this activation pattern is reversed as one would expect given that impulse vector should be directed in the opposite direction.

Main analyses

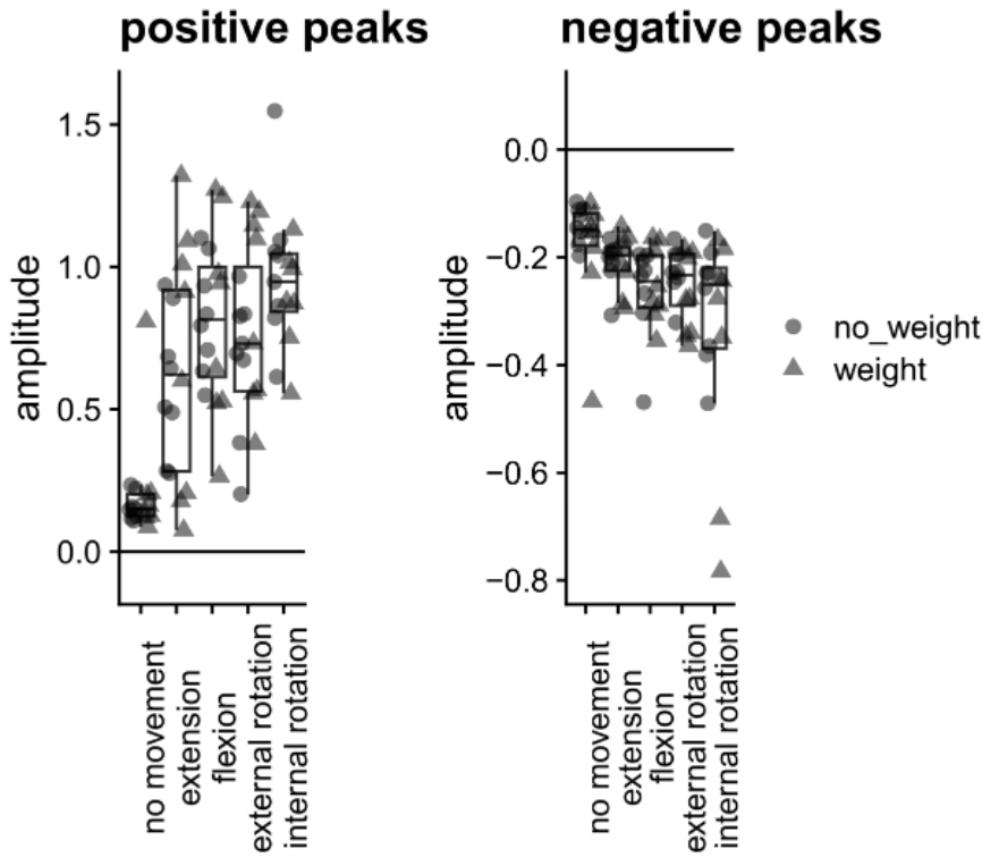
For the current pre-registration and pilot analyses we focus on the vocalization condition, ignoring the expiration baseline conditions as they are of secondary interest at this point of our inquiry.

Do different movement conditions have different effects on vocalization amplitude?

From inspecting the trajectories in Figure 5, and the descriptive exploratory analyses above, we derive that internal rotation of the arm seems to induce a negative acoustic peak around movement onset, which is followed by a positive peak. Furthermore, we observe that all other movements primarily have a positive peak. Possibly, the internal rotation also yields a negative acoustic amplitude peak slightly before peak muscle activity. A straightforward test of whether the acoustic amplitude envelope has positive or negative peaks is to assess differences in peaks in the vocalization conditions per movement (and weight) condition (see Figure 6).

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Figure 6. Effects of movement condition on positive and negative peaks in the acoustic amplitude



Note. On the left panel the positive peaks in the amplitude envelope during the different movement conditions is shown. The right panel shows the negative peaks (hence the negative values; note, in the modeling we will take absolute values). Relative to vocalization without movement, there are especially positive, but also more negative peaks in the amplitude envelope for the different movement conditions.

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Table 4. Effects of weight and movement condition on magnitude positive peaks in amplitude envelope

	Estimate	SE	df	t-value	p-value
Intercept (no movement/no weight)	0.164	0.101	3.877	1.630	.181
vs. weight	0.062	0.063	73.000	0.985	.328
vs. extension	0.436	0.100	73.000	4.357	<.001
vs. flexion	0.618	0.100	73.000	6.172	<.001
vs. internal r.	0.568	0.100	73.000	5.672	<.001
vs. external r.	0.794	0.100	73.000	7.928	<.001

We first modeled with a mixed linear regression the variation in *positive* peaks in the amplitude envelope (using R-package `lme4`), with participant as random intercept (for more complex random slope models did not converge). A model with *weight* and *movement condition* explained more variance than a base model predicting the overall mean, Change in χ^2 (5.00) = 53.68, $p < .001$. The model coefficients are given in Table 4. Wrist weight has a positive but not statistically reliable effect on acoustic amplitude. Further, all movements (extension, flexion, internal rotation, external rotation) lead to statistically reliable increases in positive peaks of the acoustic amplitude relative to the no movement condition (with flexion and external rotation leading to more extreme effects).

Secondly, we modeled in a similar way the negative peaks in the acoustic amplitude envelope, and found that a model with *weight* and *movement condition* explained more variance than a base model predicting the overall mean, Change in χ^2 (5.00) = 21.99, $p < .001$. Model coefficients are shown in Table 5. We find that some movement conditions (flexion, internal rotation, and especially external rotation) had larger acoustic magnitude negative peaks relative to no movement. Again, no reliable effect of weight condition was found, nor did the extension movement lead to negative peaks relative to the no movement condition.

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Table 5. Effects of weight and movement condition on magnitude negative peaks in amplitude envelope

	Estimate	SE	df	t-value	p-value
Intercept (no movement/no weight)	0.149	0.028	74	5.290	.000
vs. weight	0.034	0.023	74	1.494	.139
vs. extension	0.043	0.036	74	1.192	.237
vs. flexion	0.090	0.036	74	2.484	.015
vs. internal r.	0.084	0.036	74	2.298	.024
vs. external r.	0.159	0.036	74	4.374	<.001

Note. We have absolutized the values of negative peaks, such that positive effects of some condition means higher magnitude negative peaks (i.e., more negative peaks).

Power analyses. We use ‘mixedpower’ package (Kumle et al., 2021) to determine the estimated power to detect an effect if present, based on data driven simulations from the pilot data. Table 6 and 7 show the estimated power for a simulated number of participants. With about 15 to 20 participants we reach between 80% to 100% power for all fixed effects.

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Table 6. Power analyses for effects of movement and weight condition on positive peak in amplitude envelope

10	15	20	25	30	35	40	effect
0.578	0.764	0.878	0.932	0.97	0.986	0.994	Intercept (no movement/no weight)
1.000	1.000	1.000	1.000	1.00	1.000	1.000	vs. weight
1.000	1.000	1.000	1.000	1.00	1.000	1.000	vs. extension
1.000	1.000	1.000	1.000	1.00	1.000	1.000	vs. flexion
1.000	1.000	1.000	1.000	1.00	1.000	1.000	vs. internal r.

Notes. Columns indicate the number of simulated participants and values indicate the estimate power in percentages to detect a fixed effect, assuming there is an effect of a magnitude similar to the pilot data.

Table 7. Power analyses for effects of movement and weight condition on the negative peak in amplitude envelope

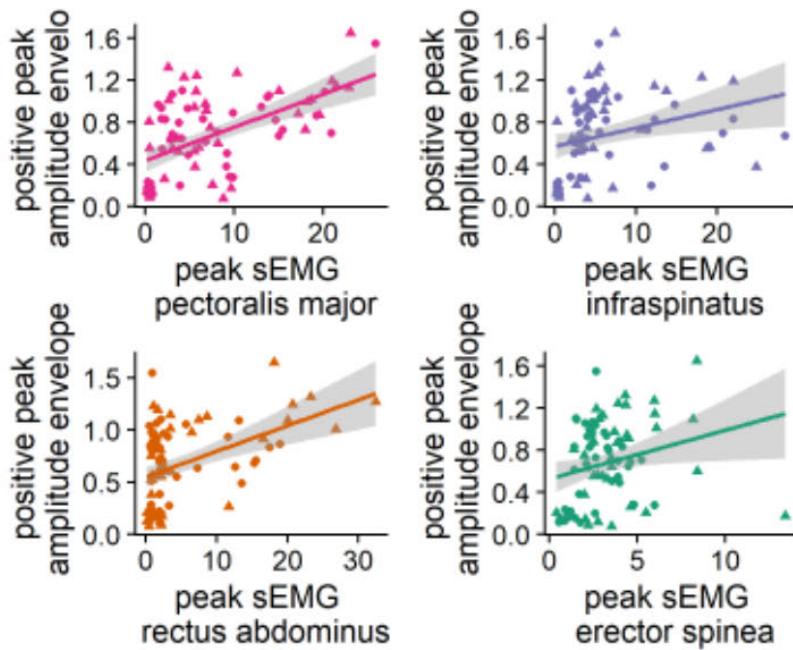
10	15	20	25	30	35	40	effect
0.908	0.990	0.998	1.000	1.000	1.000	1.000	Intercept (no movement/no weight)
0.760	0.898	0.972	0.986	0.994	1.000	1.000	vs. weight
0.998	1.000	1.000	1.000	1.000	1.000	1.000	vs. extension
0.998	1.000	1.000	1.000	1.000	1.000	1.000	vs. flexion
1.000	1.000	1.000	1.000	1.000	1.000	1.000	vs. internal r.

Is muscle activity differentially affecting amplitude of vocalization?

Since each movement and weight condition is designed to elicit different muscle recruitment patterns, we can also directly relate muscle activity peaks with the positive and negative peaks in the acoustic amplitude envelope. We again use a linear mixed regression approach to explain variance in vocal amplitude peaks based on peaks in muscle activity for the different muscles measured. Since we have already established correlations between muscle activities (see Table 2), we first assessed the VIF's between the muscle activity peaks, which yielded a maximum VIF value of 2.91. Since this is considered a low value (VIF > 5 is generally considered problematic), we can combine the different muscle activity measurements in one model to predict amplitude envelope peaks. Figure 6 shows the graphical results of these relationships.

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Figure 7. Relations between peak muscle activity and positive peaks in the acoustic amplitude envelope



A model with participant as random intercept (a more complex model with random slope for participant did not converge) the different peak muscle activities explained more variance than a base model predicting the overall mean (change in χ^2 (4.00) = 57.72, $p < .001$). The model coefficients are given in Table 8. Further, Figure 7 shows that peak EMG activity in all the muscles (but especially the rectus abdominus, a well-known expiratory muscle) leads to statistically reliable increases in positive peaks in the acoustic amplitude envelope.

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Table 8. Linear mixed regression model assessing the relation of peak muscle activity with the positive peak in the acoustic amplitude envelope

	Estimate	SE	df	t-value	p-value
Intercept	0.311	0.094	2.170	3.315	.072
erector spinae	-0.019	0.017	75.000	-1.112	.270
infraspinatus	0.009	0.005	74.052	1.825	.072
pectoralis major	0.029	0.005	74.540	5.973	< .001
rectus abdominus	0.030	0.006	50.636	5.266	< .001

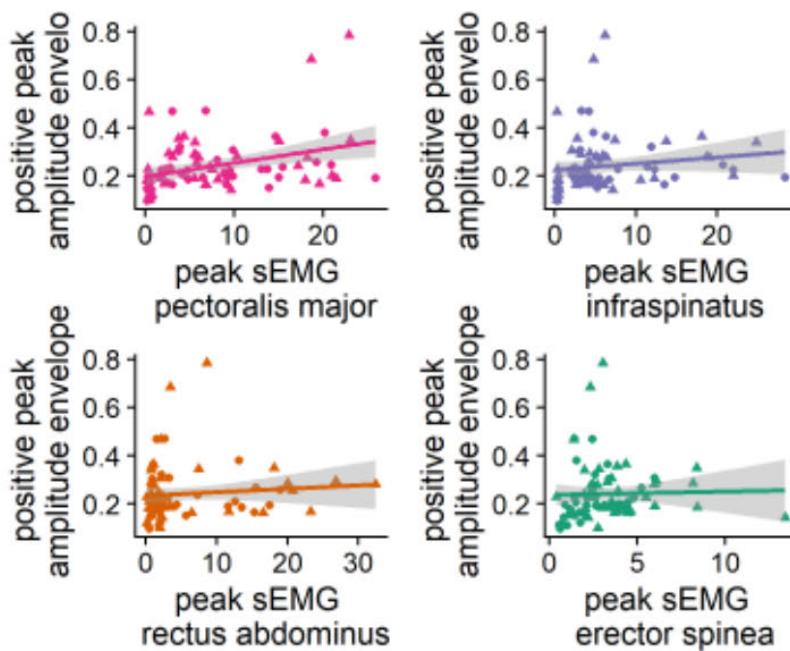
We similarly modeled the variance of the magnitude of negative peaks in the acoustic amplitude envelope with muscle activity peaks (see Figure 8 for graphical results), and we again observed that such a model performed better than a base model predicting the overall mean (change in $\chi^2 = 11.42$ (4.00), $p = 0.022$). As shown in Table 9 the pectoralis major increases in peak EMG activity led to statistically reliable increases in the acoustic magnitude of negative peaks in the amplitude envelope, while the effects of the other muscles do not reach statistical reliability.

Table 9. Linear mixed regression model assessing the relation of peak muscle activity with the positive peak in the amplitude envelope

	Estimate	SE	df	t-value	p-value
Intercept	0.204	0.026	75	7.891	< .001
erector spinae	-0.006	0.007	75	-0.861	.392
infraspinatus	0.001	0.002	75	0.342	.733
pectoralis major	0.006	0.002	75	2.968	.004
rectus abdominus	0.002	0.002	75	0.890	.376

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Figure 8. Effects of muscle activity on negative peak amplitude envelope



Note. The negative peaks in the amplitude envelope are absolutized.

Power analyses. For the model predicting positive peaks in the amplitude envelope, we already get high power for an N=10 for all predictors for determining effects of muscle activity on positive peaks in amplitude. For the negative peaks in amplitude, it is better to aim for 15 to 20 participants for the confirmatory study.

Table 9. Power analyses for effects peak muscle EMG activity on positive peak in amplitude envelope

10	15	20	25	30	35	40	effect
0.706	0.838	0.95	0.966	0.994	0.998	0.996	erector spinae
0.976	0.998	1.00	1.000	1.000	1.000	1.000	infraspinatus
1.000	1.000	1.00	1.000	1.000	1.000	1.000	pectoralis major
1.000	1.000	1.00	1.000	1.000	1.000	1.000	rectus abdominus

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Table 10. Power analyses for effects peak muscle EMG activity on positive peak in amplitude envelope

10	15	20	25	30	35	40	effect
0.542	0.638	0.780	0.888	0.908	0.950	0.982	erector spinae
0.110	0.154	0.192	0.244	0.256	0.296	0.314	infraspinatus
1.000	1.000	1.000	1.000	1.000	1.000	1.000	pectoralis major
0.460	0.656	0.786	0.874	0.918	0.966	0.970	rectus abdominus

Is activity in specific muscles related to postural stability?

Finally, we will assess which muscle activity can be related to changes in the center of pressure, which would directly confirm that gesture-speech biomechanics relates to postural stability (Pouw, Harrison, & Dixon, 2020). Figure 7 shows the graphical results. We similarly performed a linear mixed regression (with participant as random intercept) with a model containing peak EMG activity for each muscle which was regressed on the peak in change in the center of pressure (COPc). We obtained that a base model predicting the overall mean of COPc was outperformed relative to said model (change in $\chi^2 = 64.22 (4.00)$, $p < .001$). Table 11 provides the model coefficients. We find that only the postural muscles (rectus abdominus, erector spinae) indeed reliably predict the magnitude of changes in the center of pressure, while the pectoralis major and the infraspinatus do no reliably relate to the changes in center of pressure.

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Figure 7. Confirmation of postural muscle activation during changes in center of mass

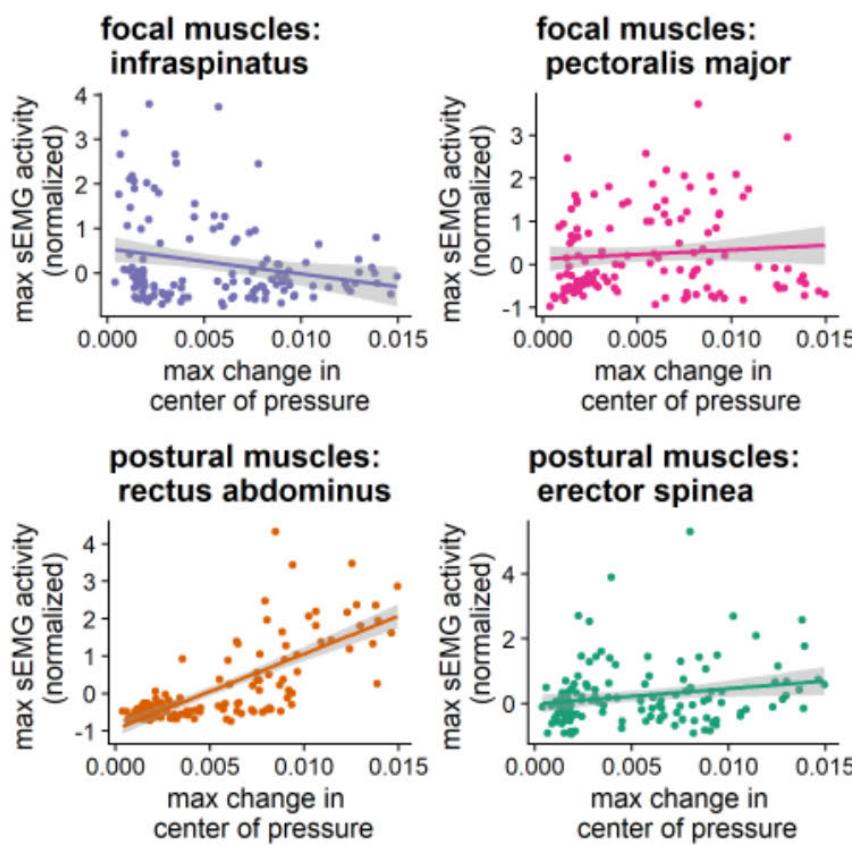


Table 11. Linear mixed regression model for predicting peak change in center of pressure based on muscle activity

	Estimate	SE	df	t-value	p-value
Intercept	-0.886	0.290	1.362	-3.051	0.146
erector spinae	0.112	0.035	74.554	3.166	0.002
infraspinatus	0.003	0.011	74.018	0.260	0.796
pectoralis major	0.016	0.010	74.208	1.613	0.111
rectus abdominus	0.074	0.012	73.406	6.219	0.000

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Power analyses. Table 12 provides a power analyses for the model assessing relationships of peak muscle activity. Similar to previous results, with about 20 participants we will reach enough power to assess fixed effects for two prime muscle targets (i.e., postural muscles).

Table 12. Power analyses for effects peak muscle EMG activity on positive peak in amplitude envelope

10	15	20	25	30	35	40	effect
0.714	0.878	0.944	0.968	0.994	1	0.998	erector spinae
0.988	1.000	1.000	1.000	1.000	1	1.000	infraspinatus
1.000	1.000	1.000	1.000	1.000	1	1.000	pectoralis major
1.000	1.000	1.000	1.000	1.000	1	1.000	rectus abdominus

Other exploratory analyses

The current pre-registration has focused on presenting a set of proof of concept analyses that a) are simple enough to support power analyses that will inform the sample size of the confirmatory study, and b) that concern the highest priority research questions in this line of research. However, we will likely perform more analyses such as time-continuous analyses of EMG and acoustic amplitude envelope (using more complex GAM modeling); Exploration of muscle activity with and without movement during vocalization versus expiration; Analyses of other acoustic parameters related to respiratory-vocal dynamics (F0). Thus, while many more analyses can be performed, we pre-register only a subset here.

Discussion

We have obtained a number of promising findings in this pilot study that support a confirmatory investigation with a larger dataset. Movement versus non-movement yields unintentional positive peaks, and to a lesser extent negative peaks, on the acoustic amplitude envelope of vocalization. Especially, for the internal rotation movement, there seems to be negative acoustic amplitude peaks associated as compared to vocalizations in the passive condition. Further analyses indeed showed that the peak activity of the internal rotator muscle (pectoralis major) reliably predicted the magnitude of the negative peak in the amplitude envelope. We can label these vocal effects as ‘unintentional’ as the task is to produce a stable vocalization output. Further, we show that all the different muscles reliably related to positive peaks in the amplitude envelope, especially the rectus abdominus, an expiratory-associated muscle. This muscle and the erector spinae are further also found to be related to postural stability, as the change in the center of pressure was reliably positively related to these postural muscle activations (in contrast to the other focal muscles). We observe small but statistically unreliable effects that a 1kg wrist weight has an effect onto the amplitude envelope (though it does affect muscle activation consistently). With a confirmatory study we will be able to confirm this effect of wrist weight, which would confirm a role of force-transfers as the mass of the moving segment is increased, therefore yielding more force per unit acceleration (Pouw, Harrison, Esteve-Gibert, et al., 2020).

If replicated, these combined results have major implications, as it is shown that upper limb motions have variable effects on the amplitude envelope depending on their exact trajectory and the underlying muscle activations, and the degree to which it affects postural stability. Consider that humans manually gesture when they speak. These gestures take varied forms, as they often depict and symbolize things in the context of speaking. Yet, it is observed that even when they take up these representational functions, they also still synchronize with the prosodic modulations of speaking (Im & Baumann, 2020; Pouw & Dixon, 2019). The current study, implies that there are complex interactions between the voice and these varied ways of gesturing. A complexity that comes with opportunities for communication, as some gestures are more aligned with certain vocal actions than others (Pouw & Fuchs, 2022).

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That some muscle activity such as the pectoralis major seems to contribute to both the magnitude of the positive and negative peaks in the amplitude envelope could have to do with the activation and release effects. Such that activation of the pectoralis leads to a negative perturbation, and the release of activity, leads to a positive perturbation. Or it relates to different effects at different joint positions (Turvey, 1982). In a further confirmatory study, we will need to look deeper into these effects by assessing the temporal ordering of the vocal, postural, and muscle peaks.

To conclude, the current pre-registration and pilot analyses provide a solid basis for a larger scale confirmatory study. It promises to show that upper limb movements affect vocalizations in complex but predictable ways.

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