nature portfolio

Peer Review File

Stretchable surface electromyography electrode array patch for tendon location and muscle injury prevention

Reviewers' comments:

Reviewer #1 (Remarks to the Author):

Yang et al. reports the fabrication of stretchable surface electromyography electrode array patch enabled by the combination of an adhesive dry electrode array and metal-polymer electrode array patch. The resulting electrode array shows long-term application capability with good durability and biocompatibility. The authors demonstrated the electrode array by the application of tendon location and muscle injury prevention. This study is well organized and the performance of the electrode array are well characterized. The results are interesting, which should be impacts on the design of dry electrodes and the development of surface electromyography. Therefore, I would like recommend the publication of this work by minor revision.

1)The mechanical properties of the electrodes are important for the practical application as dry electrodes for skin adhesion. It is suggested to provides some details characterization of the electrodes, such as modulus, stretchability, and reversibility.

2)The resolution of figures needs to be improved.

3)The authors claimed that "As a result, how to fabricate a conformal, adhesive and robust dry electrode becomes an issue to address" in the introduction. Actually, to address this issue, some recent work (cited as Ref. 3) based on the supramolecular solvent (β-cyclodextrin and citric acid), PVA and PEDOT:PSS shows soft and conformal adhesive properties as a dry electrode for the monitoring of physiological electric signals as well as flexible electronic devices. Therefore, it is suggested to provide some in depth discussion for this issue in the introduction part.

Reviewer #2 (Remarks to the Author):

This manuscript reports a stretchable surface electromyography electrode array patch, which was made by integrating liquid metal based stretchable circuit with PEDOT:PSS based soft dry electrodes. The authors first carefully checked the performance of the electrode and then showed its potential applications in predicting muscle-tendon junction location and muscle injury prevention. The research topic is interesting and the work presents some impressive results. Here are some comments which should be addressed before its consideration for publication:

1. The PPT dry electrode, as a highlight of the paper, is critically important to the recording of sEMG. However, it seems that the same material with similar recipe has already been reported and used for epidermal biopotential measurement in the literature [1]. What's the novelty and challenge of the PPT dry electrode in this paper?

2. For Figure 6, the sEMG signal pattern is strongly and frequently influenced by the innervation zone [2, 3], in this case, neither the RMS or the mean frequency could be monotonous. How to predict the muscle-tendon junction location?

3. Why is value of 0.5 suitable as the muscle-tendon junction position, please explain or add related reference.

4. The inter-electrode distance could be largely changed during the isometric task from flexion and extension due to the deformation of the sEMG array patch adhered on muscle, how to define its effect?

5. For Figure 4g, we agree that decreasing median frequencies indicated fatigue of the muscle. However, the authors claimed the decreasing value of slopes of median frequencies indicated the fact that muscle became more fatigued. If so, the slopes of median frequencies during each task (30 s) should not be a constant since the muscle is getting fatigue as the time goes. Please have a check and related reference are needed.

6. As shown in Figure 3d, TPP electrodes can work superbly with SNR level above 20 dB for almost 5 days, and then get worse. Please give the reason that decreases the SNR of PPT electrodes for long-term measurement up to 5 days.

7. How about the adhesion, skin-electrode impedance and SNR of the sEMG electrode array patch on the skin when after, e.g., 200 times, movements (compress or stretch)?

8. Could be the sEMG electrode array patch be used repetitively?

9. Please specific the thickness of each layer of the sEMG electrode array patch.

10. Bipolar recording was used for single-channel TPP electrode (Figure 4e) and unipolar recording was used for MEAP, why and what's the different?

Reference

[1]. Cao, J. et al. Stretchable and Self-Adhesive PEDOT:PSS Blend with High Sweat Tolerance as Conformal Biopotential Dry Electrodes. ACS Appl. Mater. Interfaces 14, 39159–39171 (2022).

[2]. Farina D, Madeleine P, Graven-Nielsen T, et al. Standardising surface electromyogram recordings for assessment of activity and fatigue in the human upper trapezius muscle[J]. European journal of applied physiology, 2002, 86(6): 469-478.

[3]. Beretta Piccoli M, Rainoldi A, Heitz C, et al. Innervation zone locations in 43 superficial muscles: toward a standardization of electrode positioning[J]. Muscle & nerve, 2014, 49(3): 413-421.

Reviewer #3 (Remarks to the Author):

The manuscript describes a surface electromyography electrode, which is novel, as the authors state, because it is characterised by the fact that it adheres adhesively to the skin surface, is stretchable and forms an array (see abstract). This claim by itself shows that the authors seem to be unfamiliar with the state-of-the-art in surface electromyography (sEMG). Adhesive sEMG electrodes that adhere independently to the surface of the skin have been available for several years. The manuscript does not comment on this, nor does it compare the supposedly so good new electrode with it. Instead, an unspecified Ag/AgCl electrode is used for comparison, which, as can be seen from the figures, does not correspond to the standard for sEMG electrodes. It is therefore doubtful to what the electrode introduced in the manuscript is comparedt to and how meaningful this comparison is. Electrode arrays that adhere to the skin surface for long periods of time have also been described since the 1990s and are now commercially available. There is no reference to this in the manuscript either, nor is the introduced electrode compared to them.

This leaves the property of stretchability, which according to the authors should improve the quality of sEMG signals. The advantages and disadvantages of stretchable electrode arrays have been debated among sEMG experts for many years. The problem is, that the interelectrode distance changes when the array is stretched. This affects the frequency spectrum of the sEMG signal in the case of a bipolar lead. Investigations in the frequency domain, as suggested by the authors for fatigue detection, are therefore not valid for non-isometric contractions, as it is not possible to exclude beyond doubt that a measured change in the frequency domain is not due to a change in the electrode distance. Stretchable electrode arrays are therefore fundamentally unsuitable for such applications.

This brings me to another problem concerning the manuscript. The manuscript is full of claims - often in the superlative - about signal quality and possible applications of the described electrode, which are not statistically proven. They seem to be the purely subjective perceptions of the authors. This becomes particularly clear in Fig. 2 k, in which a signal with a motion artefact, which occur from time to time but not regularly, was compared with the signal detected with the introduced electrode. The manuscript does not describe whether and if so how repeat measurements were carried out and how these were statistically evaluated to substantiate the statements made.

Fig. 4 a and e shows another problem that arises when characterising the quality of the novel electrodes. The electrodes of the devices used for comparison are not located in the same position as the novel electrodes. Rather, the comparison signals are derived at less favourable positions, which has a negative influence on the signal amplitude, the SNR and the frequency spectrum. An objective comparison between the two devices is not possible under these conditions.

Finally, a comment on electrode arrays. The use of electrode arrays has been known for a long time under the pseudonym High Density sEMG (HDsEMG) and is widely used in different research questions. The method is called sEMG imaging and the "heat maps" shown in Fig. 5 g and h are called "muscle activity maps" in the literature. The fact that HDsEMG is suitable for localising anatomical structures such as the neuromuscular junction or tendon insertion has been known since the 1990s and has been studied in a number of different investigations. In connection with fatigue and pain, a change in the spatial distribution of the activity of the muscle has already been demonstrated, as well as a change in the spectrum of the signal. This fundamental work is not mentioned anywhere in the manuscript. Rather, the impression is given that such investigations are only made possible by the new type of electrode.

Response to reviewers for the manuscript (NCOMMS-22-46103A-Z)

Reviewer #1 (Remarks to the Author):

Yang et al. reports the fabrication of stretchable surface electromyography electrode array patch enabled by the combination of an adhesive dry electrode array and metal-polymer electrode array patch. The resulting electrode array shows long-term application capability with good durability and biocompatibility. The authors demonstrated the electrode array by the application of tendon location and muscle injury prevention. This study is well organized and the performance of the electrode array are well characterized. The results are interesting, which should be impacts on the design of dry electrodes and the development of surface electromyography. Therefore, I would like recommend the publication of this work by minor revision.

Our response: We appreciate the reviewer taking the time to carefully read our manuscript and provide such excellent feedback.

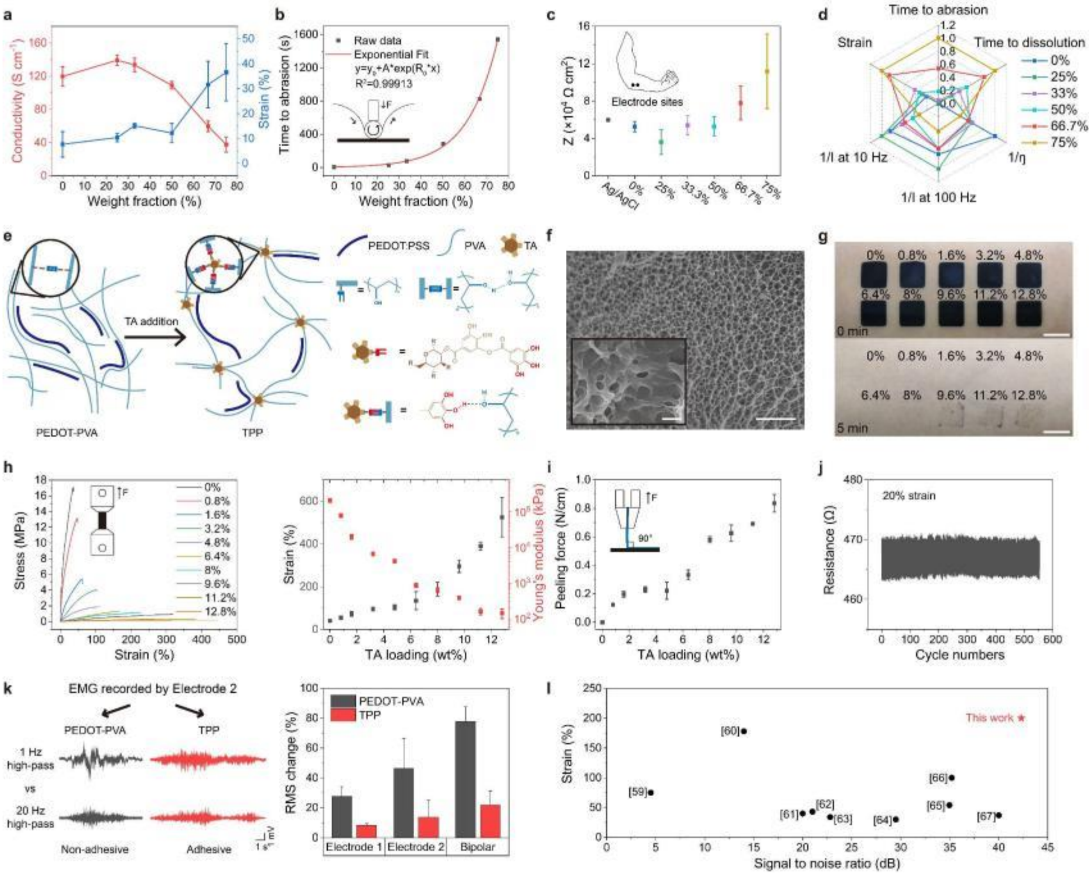
1)The mechanical properties of the electrodes are important for the practical application as dry electrodes for skin adhesion. It is suggested to provides some details characterization of the electrodes, such as modulus, stretchability, and reversibility.

Our response: We appreciate the reviewer pointing out the lack of mechanical characterizations. Following the reviewer's advice, we conducted a few studies, including tensile testing and repeated stretch measures. We agree that these findings are significant, so we include them in Fig. 2.

Meanwhile, we transfer the original Fig. 2j to Supplementary Fig. 7.

Our modifications on Page 8:

Lower Young’s modulus gives a better compliance and stretchability to the film, which is are vital to the conformal adhesion between electrodes and skin[58]. This was also proved by the results of tensile and peeling tests of TPP films with increasing concentration of TA (Fig. 2h, i). Such observations helped us use the final weight concentration of TA at 8%, which makes the film soft and adhesive but not easy to tear. This TPP film shows elongation at break of 188%, Young’s modulus of 644 kPa and adhesive forces of 0.58 N/cm on the skin. Once the concentration of the constituents of the TPP solution was determined, each constituent’s indispensability was verified by the changes in conductivity, stretchability, and adhesiveness of the electrode (Supplementary Fig. 7). Meanwhile, such TPP film showed good repeatability after being stretched to a strain of 20% for 1000 cycles (Fig. 2j).



50

51 ...Scale bar: 4 μm; inset: 1 μm.

h Tensile stress–strain curves, strain and Young’s modulus of Strain of TPP films.

i Peeling force of TPP films on the skin.

j Real-time monitoring of the TPP film by stretching the film from a strain of 0 to 20% for about 500 cycles.

k EMG signals recorded by PEDOT-PVA and TPP electrodes. Electrode...

See Page 32: ‘The tensile testing was performed by a universal testing system (Instron 68TM-5, USA), size of PEDOT-PVA and TPP films was 30 mm long and 10 mm wide. The stroke speed of the measurement was 0.5 mm min^{-1}.’

2)The resolution of figures needs to be improved.

Our response: We thank the reviewer for the requested change in resolution. We provide updated PDF document with higher resolution. We have Tag Image File Format for each figure if this manuscript gets published.

3)The authors claimed that "As a result, how to fabricate a conformal, adhesive and robust dry electrode becomes an issue to address" in the introduction. Actually, to address this issue, some recent work (cited as Ref. 3) based on the supramolecular solvent (β-cyclodextrin and citric acid), PVA and PEDOT:PSS shows soft and conformal adhesive properties as a dry electrode for the monitoring of physiological electric signals as well as flexible electronic devices. Therefore, it is suggested to provide some in depth discussion for this issue in the introduction part.

Our response: We value the reviewer's careful and thoughtful feedback on our manuscript. This statement prompted us to consider explaining material choices for flexible electronics. Conductivity is the basis, which allows materials to be classified into four types: metal, carbon materials, hydrogels, and conductive polymers. Normal metal (except liquid metal) cannot be stretched unless modified to special structure, and it is extremely difficult to impart adhesiveness on the metal itself; carbon materials, such as carbon nanotubes and graphene, have a high Young's modulus, making them unsuitable for bioelectronics (Matter (2022) 5, 1104-1136). In comparison, employing hydrogel and conductive polymers that can be fine-tuned become advantageous tactics since researchers may provide them specific functionalities depending on the application scenarios. Unfortunately, it is quite difficult to create a material that is ideal in every way. Sometimes improving one property of a material implies sacrificing another. For example, hydrogel has a lot of water but dehydrates quickly; when the conductive polymer is more stretchable, it becomes less conductive. This is also mentioned in Ref. 3 (Nature Communications (2022) 13:358), which states: '*Taking into account the compromise in mechanical flexibility, conductivity, and interface adhesion (in subsequent discussions), SACPs with PEDOT:PSS mass ratio of 3.6% presented suitable mechanical property (modulus of 401.9 kPa) and conductivity (3.79 S/cm) meet the requirements of bioelectrode.*'

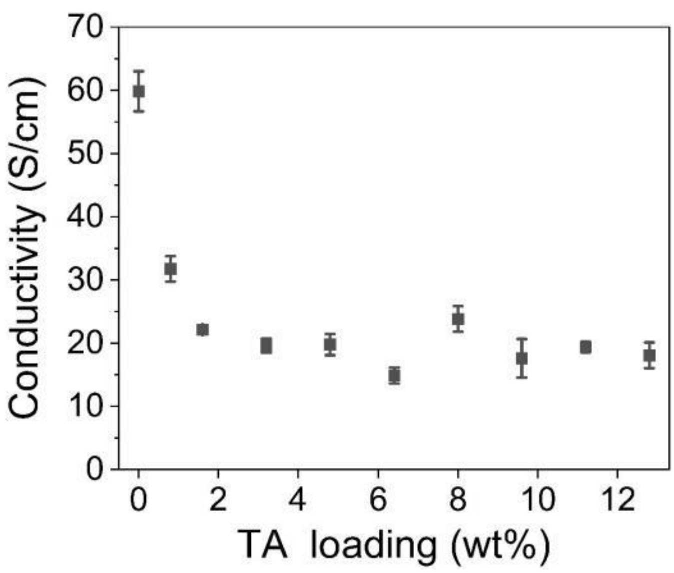
We also need to point out that Ref. 3 is an excellent contribution to the field of dry electrode, but our work has a slightly different focus to Ref. 3. The main point of Ref. 3 is to show the potential of SACPs for future bioelectronic devices, for example, that making a better tool to visualize EMG. The focus of our work is to provide an array that can monitor EMG over time to provide detailed information from different features of muscles. Additionally, the applications of SACPs in Fig. 5 and 6 of Ref. 3 didn't mention an array, which increased our appreciation for the array design of our MEAP because commercial gel electrodes cannot accomplish the same recording sites in the same area as MEAP. This difficulty is simply solved by using patternable liquid metal circuitry. We must underline this originality once again.

Considering the above, we add the paragraph on Page 4: 'As a result, how to fabricate a conformal, adhesive and robust dry electrode becomes an issue to address. Dry electrodes force the material to be classified into three types: metal, carbon materials, and conductive polymers. Conventional metal and carbon materials have exceedingly high Young's modulus, which must be fabricated into micro-/nano-structures using complicated procedures for flexible bioelectronics. In addition, employing conductive polymers that can be variably tuned becomes a more advantageous method since researchers may provide them specific functionalities depending on the application scenarios.'

We also add results of conductivity of TPP films with different TA loadings in Supplementary Fig. 9:

See page 8: ‘The conductivity and electrode-skin impedance of TPP film were also examined (Supplementary Fig. 9, 10).’

We thank the reviewer once more, for the thoughtful assessment and comments. We appreciated the chance to respond to the reviewer's suggestions in this revised



manuscript because we considered their advice to be really insightful.

Reviewer #2 (Remarks to the Author):

This manuscript reports a stretchable surface electromyography electrode array patch, which was made by integrating liquid metal based stretchable circuit with PEDOT:PSS based soft dry electrodes. The authors first carefully checked the performance of the electrode and then showed its potential applications in predicting muscle-tendon junction location and muscle injury prevention. The research topic is interesting and the work presents some impressive results. Here are some comments which should be addressed before its consideration for publication:

Our response: We value the reviewer's time spent reading our manuscript thoroughly and providing generally encouraging feedback. We are quite appreciative that the reviewer found the material to be innovative and recognized its potential for use in sports health and injury prevention.

1. The PPT dry electrode, as a highlight of the paper, is critically important to the recording of sEMG. However, it seems that the same material with similar recipe has already been reported and used for epidermal biopotential measurement in the literature [1]. What's the novelty and challenge of the PPT dry electrode in this paper?

Our response: We appreciate the reviewer's comment. Prior to submission we did find this article (Ref. 1), but it is important that the differences are carefully noted. The composition of ours are distinctly different from those mentioned in the article, resulting in our electrodes being very different in corresponding properties. A comparison table is presented below to show the difference in properties clearly and this was mentioned in Supplementary Table 1 in the original manuscript.

	Materials for substrates	Multi-channel?	Young's modulus	Strain	The adhesiveness of electrode (N/cm)	The smallest area of the electrode (mm^2)	Electrode-skin impedance at 100 Hz ($K\Omega*cm^2$)	Long-term test (Hour)	RMS of Noise (μV)	Signal-to-noise ratio (dB)
Ref.1	N/A	No	18.3 MPa	54%	0.28	16	100	N/A	11.8	34.96
This work	**PDMS**	**Yes**	**645 kPa**	**188%**	**0.58**	**0.8**	**80**	**120**	**1.0**	**42.3±0.7**

Our dry electrode has a far lower Young's modulus than theirs, which makes it softer, stickier, and more stretchable. The recorded EMG signal demonstrates how each of these elements helps electrodes adhere to the skin more effectively. Our electrodes' baseline noise decreased by a factor of ~10 than theirs, which ultimately results in a greater SNR. In order to investigate muscle loads, exhaustion, and tendon displacements, high-quality recording is essential. Our work has more potential for numerous applications due to its long-term stability (a lifetime of 5 days). More importantly, most of these electrodes—including Ref. 1—only perform the same function as Ag/AgCl hydrogel electrodes. The more literature we study, the more we believe that our liquid metal circuits in the patch is significantly distinct and superior. The liquid metal circuits enable multiple electrodes to record simultaneously over extended periods of time. We can create an array patch using this stretchable circuit, which further allows us to map muscle activity and locate muscle-tendon junctions. The fact that these measurements cannot be performed using Ag/AgCl hydrogel electrodes must be emphasized. In this paper, we think the dry electrode array is much more unique than the dry electrode itself.

To better clarify our novelty, we added following sentence in Discussion. See page 29: ‘However, commercial hydrogel electrodes cannot accomplish the same recording sites in the same area as MEAP. This difficulty is simply solved by using patternable liquid metal circuitry. We must underline this originality once again.’

2. For Figure 6, the sEMG signal pattern is strongly and frequently influenced by the innervation zone [2, 3], in this case, neither the RMS or the mean frequency could be monotonous. How to predict the muscle-tendon junction location?

Our response: We appreciate the reviewer's meticulous and in-depth work. It is accurate to say that ‘the innervation zone often and substantially influences the sEMG signal pattern; hence, neither the RMS nor the mean frequency could be monotonous.’ Yet only when the sEMG is captured using a bipolar (single differential) arrangement can this conclusion be made. As stated in ‘*Part 2.1*’ of Ref. 4 ‘*... can be detected using the monopolar or single differential (SD, bipolar) technique, ...*’ the ‘bipolar’ and ‘single differential’ montage are the same, but distinct from the "monopolar.’ On closer examination of the 2 articles the reviewer provided, it was found that the signals were identified in single differential mode in

A) the ‘*Method*’ section of Ref. 2 that ‘*The signals were detected in single differential mode to minimize line interference, ...*’;

B) in the ‘*Equipment*’ section of Ref. 3 that ‘*... or 2.5 mm (silver pins 1 mm long, 1 mm diameter) (LISiN and OT Bioelettronica, Turin, Italy) in single differential configuration.*’

In addition, as seen in Fig. 4 in Ref. 4, the amplitude of EMG recorded in monopolar mode, decreases from the innervation zone to the tendon area, which is monotonous. Based on this, we can identify the muscle-tendon junction location. We regret for using the word ‘unipolar’ instead of ‘monopolar’ in the paper, which may have caused some misunderstanding. Thus, on page 34, we make the following changes:

See Page 34: ‘Bipolar recording was used for single-channel TPP electrode and monopolar recording was used for MEAP to obtain sEMG signals.'

3. Why is value of 0.5 suitable as the muscle-tendon junction position, please explain or add related reference.

Our response: We appreciate the reviewer bringing this to our attention. We are sorry that we cannot locate any relevant references because we are the first team to do muscle-tendon junction localization using a sEMG electrode array, but this value of 0.5 was born with repeated verifications. When we used the MEAP to record on the biceps, we discovered that various channels responded differently in mean frequency (see Supplementary Fig. 16), and they consistently maintained a monotonous order from tendon to muscle. We can plainly discern the junction site and its relative position with our array using ultrasound images (the gold standard for tendon monitoring) (see Fig. 7b). After comparing the results, we discovered that the junction location was usually near to 0.5 after normalization (see Supplementary video 5). In other words, we always saw the junction between two channels with normalized values closer to 0.5 (see Fig. 7c). We further confirmed this rule by palpating the biceps and Achilles tendon directly, and eventually established that 0.5 was the best value to discern the junction point. After that, we tested this rule with palpation on subjects 2 and 3. The outcomes were likewise comparable. We accept that this method of locating may not be as exact as magnetic resonance imaging or ultrasound imaging, but the convenience provided by our array should be underscored. Instead of making this document excessively confusing, we decided to examine and summarize a mature method or accuracy improvements in another study.

To clarify, we amended the phrase on page 22: 'According to the results of ultrasound image, we found the junction location was always close to the value of 0.5 after normalization. We defined the channel with value of 0.5 is suitable as the muscle-tendon junction position (Supplementary Text 1). This observation was also verified by palpation on the biceps brachii and the Achilles tendon.'

4. The inter-electrode distance could be largely changed during the isometric task from flexion and extension due to the deformation of the sEMG array patch adhered on muscle, how to define its effect?

Our response: We appreciate the reviewer bringing this IED problem to our attention. RMS and frequency can fluctuate as the IED between two electrodes changes, especially in bipolar montage recording. Nevertheless, the recording setups are monopolar for the majority of applications employing MEAP in this publication, including identification of muscle loading, muscle exhaustion, and muscular activity map. Signals from each electrode are unrelated to one another. The sole application that may be affected is the location of the muscle-tendon junction, because the IEDs are not constant during muscle activity, which may produce an inaccuracy in the quantitative value of tendon displacements. However, we must underline that changing the IEDs will not affect the detection of junctions using our normalizing technique. Because the junction should always be located between two neighboring channels, we may utilize the value of 0.5 to establish which two channels are involved. In terms of the numerical value of tendon displacement, we have previously studied this issue and modified computation to reduce the effect as much as feasible. We added the demonstration in Supplementary Text 2.

See page 22: 'We also improved calculation to make the influence of skin deformation as low as possible (Supplementary Text 2).'

Supplementary Text 2:

We measured the IEDs between neighboring channels at the muscle-tendon junction of the biceps distal tendon.

Flexion degree	IED_{21-17} (mm)	IED_{17-13} (mm)	IED_{13-9} (mm)	IED_{9-5} (mm)	IED_{5-1} (mm)
30°	15 mm	15 mm	15 mm	15 mm	15 mm
0°	15 mm	15 mm	16 mm	18 mm	18 mm
110°	15 mm	15 mm	14 mm	13 mm	12 mm

Since TPP electrodes are sticky and can adhere securely to the skin, the IED change is the same as skin deformation between two nearby electrodes. We evaluated IEDs in three distinct arm states: full extension (0°), full flexion (110°), and relax (30°). When the array was bonded to the skin during arm relaxation, the IEDs were all 15 mm. We discovered that when the muscle is moving, the deformations of skin on the muscle part are not obvious.

For example, we assume the junction is right in the middle between channels 5 and 1 when the muscle is during full extension.

The calculated distance between junction and channel 21 is $D_{ce} = IED_{21-17}(30°)+IED_{17-13}(30°)+IED_{13-9}(30°)+IED_{9-5}(30°)+1/2\ IED_{5-1}(30°)$;

the realistic distance between junction and channel 21 is $D_{re} = IED_{21-17}(0°)+IED_{17-13}(0°)+IED_{13-9}(0°)+IED_{9-5}(0°)+1/2\ IED_{5-1}(0°)$ 。

When the junction is right middle between channel 13 and 17 when full flexion, the calculated distance between junction and channel 21 is $D_{cf} = IED_{21-17}(30°)+1/2\ IED_{17-13}(30°)$;

the realistic distance between junction and channel 21 is $D_{rf} = IED_{21-17}(110°)+1/2\ IED_{17-13}(110°)$.

Considering the absolute displacement D_a of channel 21 between flexion and extension in the space,

then the calculated displacement is $D_{ce}-D_{cf} + D_a = 1/2\ IED_{17-13}(30°)+IED_{13-9}(30°)+IED_{9-5}(30°)+1/2\ IED_{5-1}(30°) + D_a$;

the realistic displacement is $D_{re}-D_{rf} + D_a = 1/2\ IED_{17-13}(0°)+IED_{13-9}(0°)+IED_{9-5}(0°)+1/2\ IED_{5-1}(0°)+D_a$;

The reason we choose distance between junction and channel 21 instead of channel 1 is that we found $D_a(21) \approx 0$ mm, while $D_a(1) \approx 40$ mm.

Calculating from the muscle end can therefore reduce the effect of significant skin distortion.

Therefore, the computed displacement is 45 millimeters, but the real displacement is 50.5 millimeters. The inaccuracy is roughly 10%, which is deemed acceptable.
We did not see evident skin deformation between flexion and extension for Achilles tendon identification, hence that we chose not to include this component in that application.

Subject No.1's body fat percentage is roughly 20%, thus we don't observe many skin deformations. Nevertheless, for persons with a body fat content of less than 15% or even 10%, skin deformation can produce significant changes in IEDs and a higher risk of non-conformal electrodes peeling off. In such instance, the stretchable electrode array is more helpful and relevant, but it works best when combined with a strain sensor on the patch to monitor and reduce the influence of skin deformation.
Similarly, we do not think these solutions should be discussed in this document to ensure this manuscript is not too disorganized.

5. For Figure 4g, we agree that decreasing median frequencies indicated fatigue of the muscle. However, the authors claimed the decreasing value of slopes of median frequencies indicated the fact that muscle became more fatigued. If so, the slopes of median frequencies during each task (30 s) should not be a constant since the muscle is getting fatigue as the time goes. Please have a check and related reference are needed.

Our response: We appreciate the reviewer bringing up the problem of fatigue. True, the slope of median frequency does not remain constant during a task, especially if the task duration is long. As time passes, the median frequency decline will be slower because the source of tiredness is a reduction in the number of available motor units to be innervated. We can see that the exponential regression may very well match the median frequency reduction (See Fig. 5 of Ref. 5). The primary decline of median frequency occurs in the 30s after the task starts. The issue is that utilizing the exponential regression approach makes it difficult to compare each fatigue phase quantitatively, but using linear fitting to get slope values allows us to do so. In fact, using slope value is a common strategy to research muscle fatigue (Ref. 6).
In part 3.4 of Ref. 6, you can also see that the slopes of median frequency are lower with increasing percentage of MVC, which corresponds to our statement in the original manuscript that ‘the decreasing value of slopes recorded by TPP electrodes

also matched the fact that muscle became more fatigued after each task.’ Thanks to the research mentioned by Reviewer 2, we can improve our new experiment design by isolating the first 25 seconds of each task to undertake linear fitting to decrease error as much as feasible. We appreciate the reviewer's thoughtful and considerate comments, which will be extremely valuable for this work and future study.

See page 14: ‘To assess the ability of TPP electrodes to obtain information of frequency in the signal, Ag/AgCl and TPP electrodes were set on the same position on FCU, and the subject was asked to curl the wrist with a 5 kg dumbbell for three long periods to activate FCU (Fig. 4e). The TPP electrodes showed a little better SNR than the Ag/AgCl electrodes that they are 39.2, 37.5 and 40.5 dB for three contractions recorded by TPP electrodes and 38.9, 37.5 and 38.6 dB by Ag/AgCl electrodes. The spectrograms showed TPP electrodes can give clear frequency information just like Ag/AgCl electrodes (Fig. 4f). To compare the performances of Ag/AgCl and TPP electrodes on fatigue measurement, the subject was asked to curl the wrist 60 s for three times for each type of electrodes. Three tasks were named as flexion 1, 2 and 3 to calculate median frequency during each task (Fig. 4g). Linear fittings were made for first 25 s of each contraction, to quantify the outcome with less errors[73]. The slopes obtained by two types of electrodes both showed negative which indicated the muscle was in fatigue. This test proved the TPP electrodes can measure the muscle fatigue the same as Ag/AgCl electrodes.’

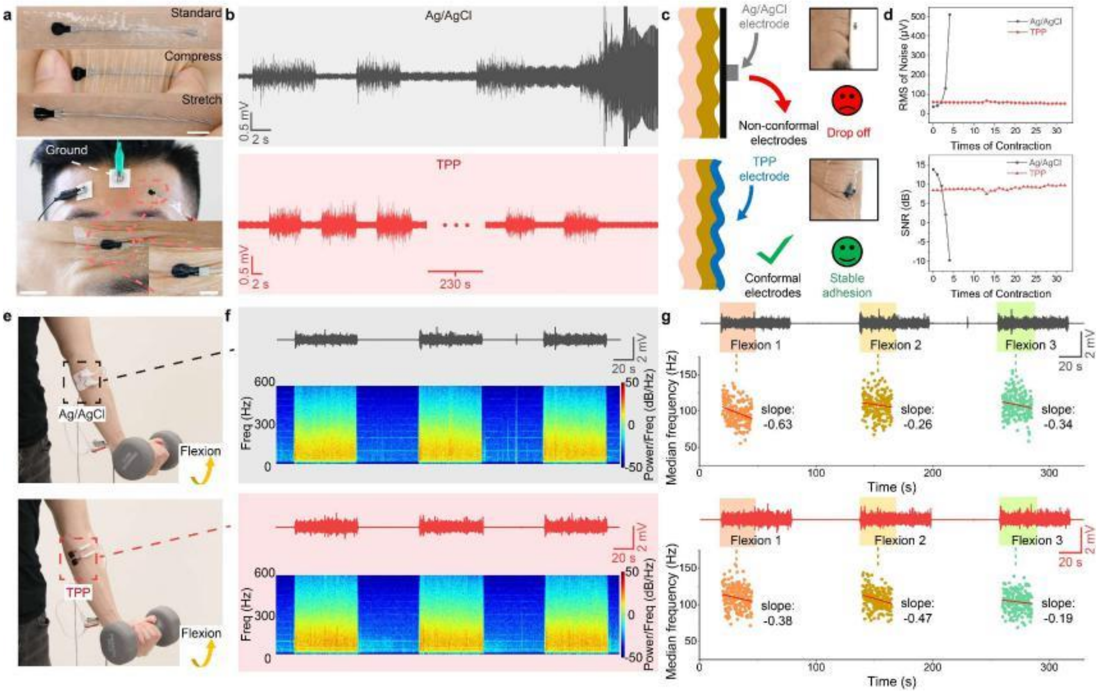


Fig. 4 Comparison of recording performances on skin between Ag/AgCl and TPP electrodes.

a Up, standard, compressing and stretching TPP electrodes on the skin. Scale bar: 1 cm; bottom, photographs of Ag/AgCl and TPP electrodes when recording sEMG of frontalis and the TPP electrode in the skin folds. Scale bar of photo at the bottom: 1 cm; bottom inset: 0.5 cm.

b sEMG signals recorded by Ag/AgCl and TPP electrodes, respectively. The subject was asked to make each contraction for 5 seconds. In the case of recording by Ag/AgCl electrodes, after four times of contraction, noises were even higher than signals.

c Schematic illustrations and lateral photos of Ag/AgCl electrode and TPP electrode on skin folds.

d Noise level and SNR recorded by two electrodes during contractions.

e Photographs of electrode configuration on FCU and contraction task. Two pairs of electrodes were attached the same position on the forearm.

f sEMG signals and spectrograms recorded by Ag/AgCl and TPP electrodes respectively.

g sEMG signals and fitting results of median frequency during flexion 1, 2 and 3 recorded by Ag/AgCl and TPP electrodes. Decreasing median frequencies indicated fatigue of the muscle.

6. As shown in Figure 3d, TPP electrodes can work superbly with SNR level above 20 dB for almost 5 days, and then get worse. Please give the reason that decreases the SNR of PPT electrodes for long-term measurement up to 5 days.

Our response: The drop in SNR (caused by an increase in baseline noise level) of commercial Ag/AgCl electrodes is caused by desiccation of conductive gels between electrodes and skin. Many factors may contribute to a rise in baseline noise level in TPP electrodes. We hypothesize that perspiration in normal life might gradually destroy TPP electrodes because sweat fat and salts cannot escape and only slowly accumulate on the TPP film. These interface changes can change the effective contact area and conductivity of the film, increasing the noise intensity. Moreover, the increase of the thickness of stratum corneum because of normal metabolism may increase the impedance between electrodes and skin during such an extended period of measurement. Nonetheless, TPP electrodes have been shown to have a substantially longer effective use period than commercial gel electrodes.

See page 11: ‘As for the increase of baseline noise, we hypothesize that perspiration in normal life might gradually the effective contact area and conductivity of the TPP film because sweat fat and salts can only slowly accumulate on the TPP film, increasing the noise intensity[68]. Moreover, the formation increase of the thickness of stratum corneum because of normal metabolism may increase the impedance between electrodes and skin during such an extended period of measurement.’

7. How about the adhesion, skin-electrode impedance and SNR of the sEMG electrode array patch on the skin when after, e.g., 200 times, movements (compress or stretch)?

Our response:

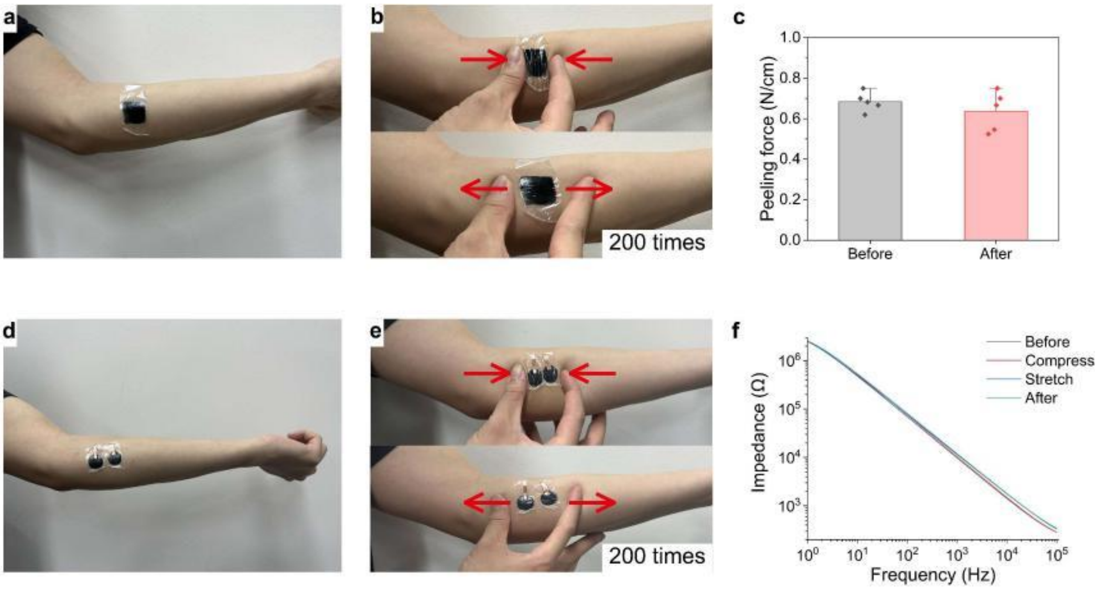
In response to this comment, we performed three tests to examine the change in adhesion, impedance, and SNR after 200 times of compression or stretching. We physically squeezed and stretched the skin with our fingertips to make the results more convincing.

See page 8: ‘Further, we found TPP electrodes showed excellent stability in adhesion, skin-electrode impedance and SNR after 200 times of compress or stretch on the skin (Supplementary Fig. 11, 12).’

Supplementary Fig. 11 The change in adherence and impedance of TPP electrodes on the skin.

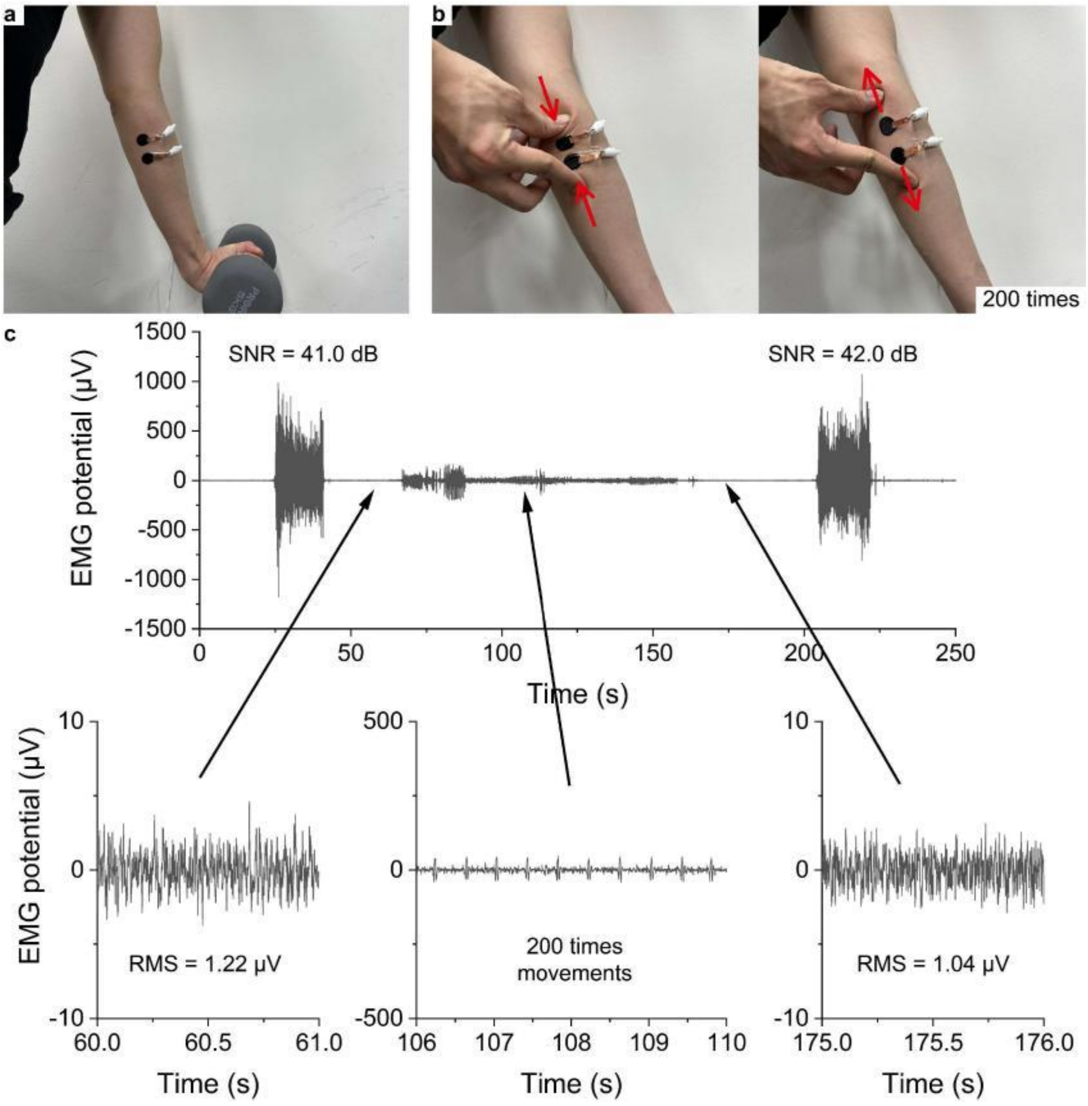
a, b images of the electrode applied to the skin and the motions made during the adhesion test.

c The peeling force of TPP electrodes off the skin before and after motions.



d, e Images of the electrode applied on the skin and the process of movements during impedance test.

f The impedance of TPP electrodes on the skin before, after motions and in the state of compressing and stretching.



Supplementary Fig. 12 SNR variation and baseline noise levels of TPP electrodes on the skin.

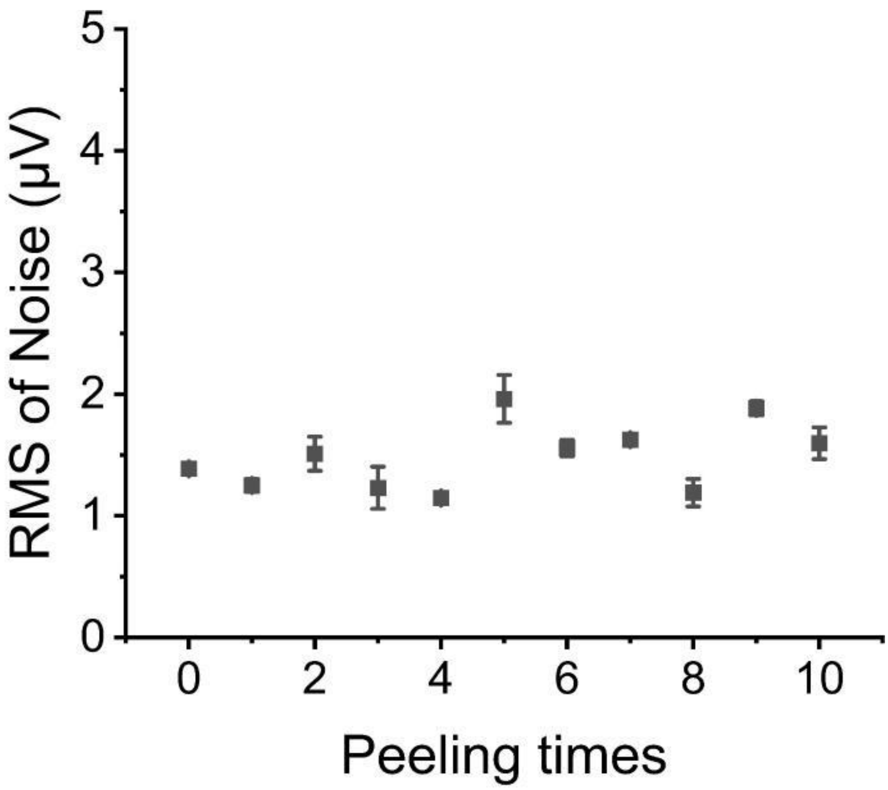
a, b Images showing the electrode applied to the skin and the motions made to compare SNR.

c Demonstration of the entire process of recording using the TPP electrodes applied on skin before and after motions. The baseline noise level was reduced to 1.04 μV from 1.22 μV which was stable even after the motions. As a result, there was little change in the SNR of signals, showing the stability of TPP electrodes on the skin even after compression or stretching.

8. Could be the sEMG electrode array patch be used repetitively?

Our response: Using our TPP electrodes on the skin, we repeatedly performed peeling-attaching studies and analyzed the noise of baseline to address this remark. Even after 10 repetitions, the noise level barely changed. PPA layer on substrate may also withstand several peelings and maintain enough adhesive force (Ref. 7). Hence, we think the sEMG electrode array patch can be reapplied repeatedly.

See page 8: 'TPP electrodes can also be used repetitively without changing the baseline noise (Supplementary Fig.13).'



Supplementary Fig. 13 The repetitive test of TPP films on the skin.

9. Please specific the thickness of each layer of the sEMG electrode array patch.

Our response: We added the following sentence to The fabrication of MEAP.

See page 31: 'The thickness of MEAPs was lower than 100 μm, typically with the substrate of 65 μm, the encapsulation layer of 25 μm and the TPP electrode of 30 μm.'

10. Bipolar recording was used for single-channel TPP electrode (Figure 4e) and unipolar recording was used for MEAP, why and what's the different?

Our response: Bipolar recording can raise the SNR level to provide clearer recordings by reducing shared noise between two electrodes. As the bipolar (single differential) design is the one that is most commonly used in the EMG area, we employed this mode for single-channel TPP electrodes to permit a direct comparison with commercial Ag/AgCl electrodes. The SNR of monopolar recording is decreased for applications employing MEAP, but it is still adequate for us to investigate information about several muscles. In order to provide more convincing spatial information, we thus create monopolar recordings to reduce the effects between each electrode. This was covered in question 4 above.

We thank the reviewer again, for the thorough review and thoughtful advice. We appreciated this precious opportunity to revise the manuscript according to the reviewer's suggestions, which are extremely valuable to the improvement of our manuscript.

Reference

[1]. Cao, J. et al. Stretchable and Self-Adhesive PEDOT:PSS Blend with High Sweat Tolerance as Conformal Biopotential Dry Electrodes. ACS Appl. Mater. Interfaces 14, 39159–39171 (2022).

[2]. Farina D, Madeleine P, Graven-Nielsen T, et al. Standardising surface electromyogram recordings for assessment of activity and fatigue in the human upper trapezius muscle[J]. European journal of applied physiology, 2002, 86(6): 469-478.

[3]. Beretta Piccoli M, Rainoldi A, Heitz C, et al. Innervation zone locations in 43 superficial muscles: toward a standardization of electrode positioning[J]. Muscle & nerve, 2014, 49(3): 413-421.

Reference for the response to the Reviewer #2

[1]. Cao, J. et al. Stretchable and Self-Adhesive PEDOT:PSS Blend with High Sweat Tolerance as Conformal Biopotential Dry Electrodes. ACS Appl. Mater. Interfaces 14, 39159–39171 (2022).

[2]. Farina D, Madeleine P, Graven-Nielsen T, et al. Standardising surface electromyogram recordings for assessment of activity and fatigue in the human upper trapezius muscle[J]. European journal of applied physiology, 2002, 86(6): 469-478.

[3]. Beretta Piccoli M, Rainoldi A, Heitz C, et al. Innervation zone locations in 43 superficial muscles: toward a standardization of electrode positioning[J]. Muscle & nerve, 2014, 49(3): 413-421.

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

[5] Merletti, R. & Roy, S. Myoelectric and mechanical manifestations of muscle fatigue in voluntary contractions. *J. Orthop. Sports Phys. Ther.* **24**, 342–353 (1996).

[6] Oliveira, A. de S. C. & Gonçalves, M. EMG amplitude and frequency parameters of muscular activity: Effect of resistance training based on electromyographic fatigue threshold. *J. Electromyogr. Kinesiol.* **19**, 295–303 (2009).

[7] Cheng, J. *et al.* Wet-Adhesive Elastomer for Liquid Metal-Based Conformal Epidermal Electronics. *Adv. Funct. Mater.* 2200444 (2022) doi:10.1002/adfm.202200444.

Reviewer #3 (Remarks to the Author):

The manuscript describes a surface electromyography electrode, which is novel, as the authors state, because it is characterised by the fact that it adheres adhesively to the skin surface, is stretchable and forms an array (see abstract). This claim by itself shows that the authors seem to be unfamiliar with the state-of-the-art in surface electromyography (sEMG). Adhesive sEMG electrodes that adhere independently to the surface of the skin have been available for several years. The manuscript does not comment on this, nor does it compare the supposedly so good new electrode with it.

Our response: We appreciate the reviewer looking through our text and bringing up any issues. We consistently study the state-of-the-art in the sEMG area in order to evaluate the uniqueness of our work objectively. The TPP electrodes were compared with other good new electrodes, and it was mentioned on page 9: 'In comparison with dry electrodes in reported literatures[59–67], TPP electrode performs better when the conformability and signal quality are evaluated (Fig. 2l, Supplementary Table 1, 2).'

Supplementary Table 1 Comparisons between dry electrodes in other literatures and this work.

	Materials for electrodes	Materials for substrates	Is it intrinsically stretchable?	Strain	The adhesiveness of electrode (N/cm)	The smallest area of the electrode (mm^2)	Electrode-skin impedance at 100 Hz ($K\Omega*cm^2$)	Long-term test (Hour)	RMS of Noise (μV)	Signal-to-noise ratio (dB)	Reference
	Ag	Polyimide	No	80%	0	16.0	12.8	11	N/A	N/A	1
	Ag-filled epoxy	Epoxy	No	N/A	0	100.0	80.0	24	~43.0	16.0	2
	Ag flakes/PDMS	PDMS	Yes	480%	0	100.0	34.0	10	~540.0	N/A	3
	Ag-polytetrafluoroethylene	Polyurethane	Yes	20%	0	600.0	N/A	N/A	~74.0	N/A	4
	Au nanoparticles	Polyimide	No	N/A	0	80.0	N/A	24	~60.0	~21.0	5
	PEDOT:PSS/Glycerol	Silk fiber	Yes	250%	N/A	314.0	~157.0	N/A	N/A	N/A	6
	PEDOT:PSS/Glycerol/Polysorbate	N/A	Yes	100%	0.013	100.0	200.0	12	N/A	35.2	7
	PEDOT:PSS/Polylactic acid	N/A	No	34%	~0.467	176.6	~35.3	N/A	~47.0	22.8	8
	PEDOT:PSS/ Poly(poly(ethylene glycol) methyl ether acrylate)	N/A	Yes	75%	0.005	400.0	N/A	N/A	~60.6	4.5	9
	PEDOT:PSS/Polyvinyl alcohol/Borax	N/A	Yes	400%	N/A	254.3	101.7	N/A	N/A	29.5±1.3	10
	WPU/Deep eutectic solvent/ Tannic acid	N/A	Yes	178%	0.125	1256.0	25	N/A	50.0	~14.0	11
	PEDOT/Waterborne polyurethane/D-sorbitol	N/A	Yes	43%	0.43	400	15	16	~25	~20	**12**
	PEDOT/Polyvinly achohol/Tannic Acid	N/A	Yes	54%	0.28	16	100	N/A	11.8	34.96	**13**
This wor	**PEDOT/Polyvinly achohol/ Tannic Acid/Liquid metal**	**PDMS**	**Yes**	**188%**	**0.58**	**0.8**	**80**	**120**	**1.0**	**42.3±0.7**	

One of these works was published in the year 2020, nine in 2021, and three in 2022. We believe that this comparison accurately captures the current state-of-the-art in sEMG electrode technology. Just 6 out of 13 studies discuss sticky electrodes, and our TPP electrodes perform the best in terms of adhesiveness, which is important to note. TPP also fared the best in terms of long-term usage and signal quality. Our claim that our TPP electrodes are currently state-of-the-art is supported by this comparison.

Nevertheless, none of these electrodes could be used to create a stretchable array, which drastically limited the number of applications that could be used in the sEMG sector. In order to evaluate the effectiveness of our work objectively, we also compared the performance of MEAP and other sEMG arrays. That part will be discussed later for another concern about electrode arrays from the reviewer. But we think the lack of detailed comparisons was the main reason for the concern from the reviewer, so we added description about the comparison with other dry electrodes.

See page 9: 'In the comparison, we also found only 6 out of 13 studies discussed sticky electrodes, and our TPP electrodes perform the best in terms of adhesiveness, which is an important contribution to its highest SNR among all dry electrodes.'

Instead, an unspecified Ag/AgCl electrode is used for comparison, which, as can be seen from the figures, does not correspond to the standard for sEMG electrodes. It is therefore doubtful to what the electrode introduced in the manuscript is comparedt to and how meaningful this comparison is.

Our response: We thank the reviewer for seeking clarification. We checked our manuscript and found the details about the Ag/AgCl electrode was only mentioned in the 'MATERIALS AND METHODS -- Impedance measurement'. The Ag/AgCl electrode (Foam Monitoring Electrode 2228, 3M, USA) we used is a standard sEMG electrode, and it has previously been reported in many articles by other researchers (Ref. 14-18). So we believe all our comparisons are reliable and convincing. But we also found the use of 2228 electrodes for sEMG just started in recent years, which might not be recognized by the sEMG field. We took the advice from the reviewer and bought Red Dot 2223 electrode from 3M for our experiments. 2223 electrodes were used in literature far longer, so we think the results should be trustworthy (Ref. 19-25). This part will also be discussed below for another concern from the reviewer.

To make the experimental details clearer, we added details in 'MATERIALS AND METHODS -- sEMG signal recording'.

See page 34: ‘Foam Monitoring 2228 electrodes were used for long-term test, flexibility test on the forehead, and Red Dot 2223 (USA, 3M) electrodes were used for fatigue tests.’

Electrode arrays that adhere to the skin surface for long periods of time have also been described since the 1990s and are now commercially available. There is no reference to this in the manuscript either, nor is the introduced electrode compared to them.

Our response: We agree with the reviewer that this comparison between commercial array (CA) and our MEAP is necessary. The material for the most popular commercial sEMG array currently available is polyimide (PI), which has the Young's modulus of 3 Gpa. Due to its characteristics, it was found that unless specific features, such as serpentine design, were introduced, the array with PI substrate cannot make a fully conformal contact with human skin (Young's modulus of 10 kPa). Also, we recorded two movies to contrast the contact effectiveness of the PI sEMG array and the MEAP on the skin (see Supplementary Video 2 and 3). Commercial array came off either on the muscle or muscle-tendon junction during the muscle action, yet the MEAP always maintained perfectly conformal contact. We also conducted sEMG recording by CA and MEAP and added the results in the manuscript.

See page 16: ‘**Comparison between MEAP and commercial sEMG array**

Electrode arrays which adhere to the skin have been developed and used in laboratory settings. The material for the most popular sEMG commercial array (CA) currently available is polyimide (PI), which has the Young's modulus of 3 GPa. Due to its characteristics, it was found that unless specific features, such as serpentine design, were introduced, the array with PI substrate cannot make a fully conformal contact with human skin (Young's modulus of 10 kPa). We recorded a movie to contrast the contact effectiveness of the PI sEMG commercial array and the MEAP on the biceps brachii (Fig. 5a, Supplementary Video 2). CA formed gaps between itself and the skin during the muscle action, yet the MEAP always maintained great contact, even though their thicknesses are both 100 μm. The differences in attachment performance were reflected directly in the sEMG signals. When CA was applied to the muscle, the poor attachment caused gaps between electrodes and the skin after the muscle contractions, resulting in the increase in baseline noise during the rest stage, which would lower the

SNR level (Fig. 5b). To demonstrate the effect of skin deformation on the recording performance, we calculated the SNR of the first and the last contractions. We recorded and displayed the SNR of each channel, and none of the CA electrodes provided SNR greater than 20 dB with the last contraction, whereas all MEAP channels provided SNR greater than 20 dB for both the first and last contractions (Fig. 5c). We believe that the conformal attachment is the determining factor in this because the SNR of the first contraction recorded by CA was adequate but significantly worsened for the last contraction. We also employed statistical analysis to quantify these outcomes (Fig. 5d). Because of the mismatch between CA and the skin, the baseline noise level of all CA electrodes rose distinctly after only one contraction. After reattachment, the baseline noise was lowered, indicating again that the mismatch between CA and skin is the cause for change. Most CA channels showed more than two-fold change in baseline noise, resulting in a significantly lower SNR. While the MEAP exhibited a much more stable noise level even after five contractions, maintaining a high SNR. We also recorded sEMG signals from muscle-tendon junctions (closer to the distal end of the biceps) with both arrays, where skin deformation was greater (Fig. 5e, Supplementary Video 3). It can be seen that the mismatch between CA and skin was significant, while the MEAP still kept perfectly conformal contact. The sEMG signals also revealed a drop-off of signal from some electrodes of CA (Fig. 5f). Recordings from the CA channels on muscle-tendon junction were significantly damaged by muscle contractions, because SNR of most channels decreased from the first contraction to the last one (Fig. 5g). MEAP on the other hand, produced stable recordings with all channels having SNR greater than 20 dB. Statistical analysis of sEMG data from CA on the distal end produced similarly unsatisfactory results, but it is worth noting that MEAPs always kept stable and excellent recording even after ten muscle contractions (Fig. 5h). These results suggest that the MEAP can record better sEMG signals than CA because of the better contact even with large skin deformation.

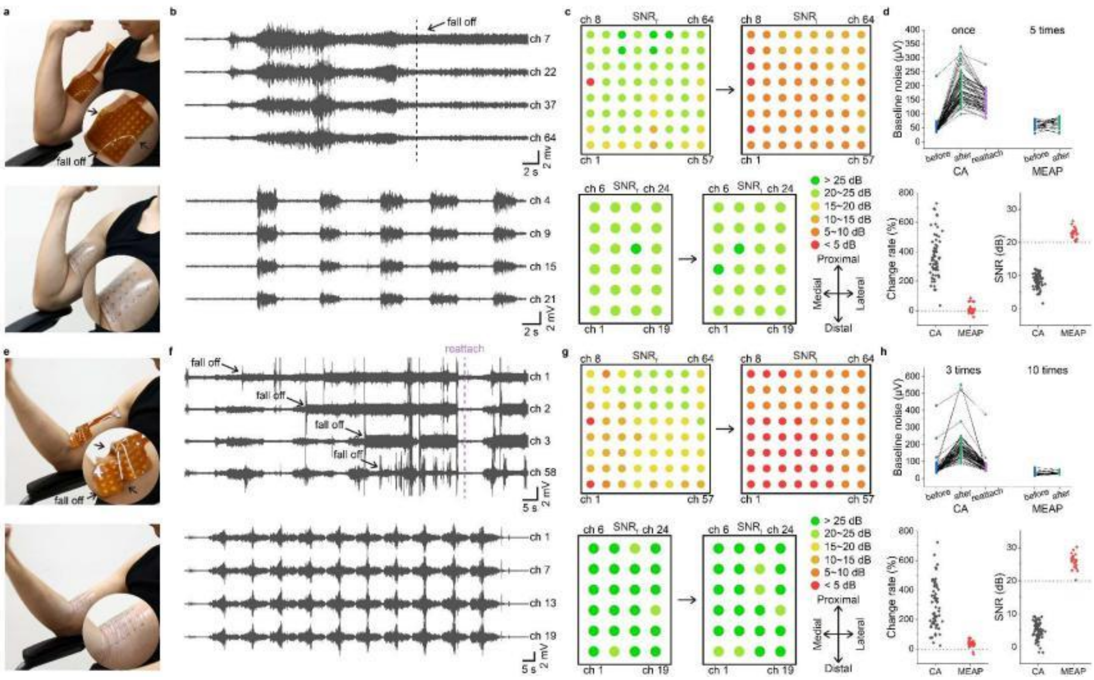


Fig. 5 The comparison of attachment performances between commercial array and MEAP.

a, e Photographs of CA and MEAP attachment, illustrating the difference when recording from muscle and muscle-tendon junction of biceps brachii.

b, f sEMG signals recorded using CA and MEAP on muscle and muscle-tendon junction of biceps brachii. Four typical channels were picked for each recording.

c, g Spatial SNR performance map for each channel of CA and MEAP for the first and last muscle contraction. SNR_f: SNR of the first contraction; SNR_l: SNR of the last contraction.

d, h Statistical analysis of performances between CA and MEAP, including baseline noise level of CA before and after one or three muscle contractions, as well as after reattachment; baseline noise level of MEAP before and after five or ten muscle contractions; baseline noise change rates before and after muscle contractions; SNR performance of the last muscle contraction recorded by each of the CA and MEAP channels.'

See page 4: 'Based on MPC circuit, a multi-channel sEMG metal-polymer electrode array patch is fabricated to achieve high-quality and high-density sEMG signals for monitoring of muscle loading and muscle fatigue, whose performance is more stable than commercial sEMG array.'

See page 34: ‘The commercial array has 64 channels Ag/AgCl electrodes on a polyimide substrate with thickness of 100 μm (Neuracle, China). The conductive gel g.GAMMAgel (G.tec, Austria) was used between commercial array and the skin.’

We can see MEAP performs better than commercial arrays since a good contact is a crucial need for dependable and steady recording. We also compared flexible arrays described in the literature in Supplemental Table 2 of the original manuscript.

Supplementary Table 2 Comparisons between sEMG arrays in other literatures and this work.

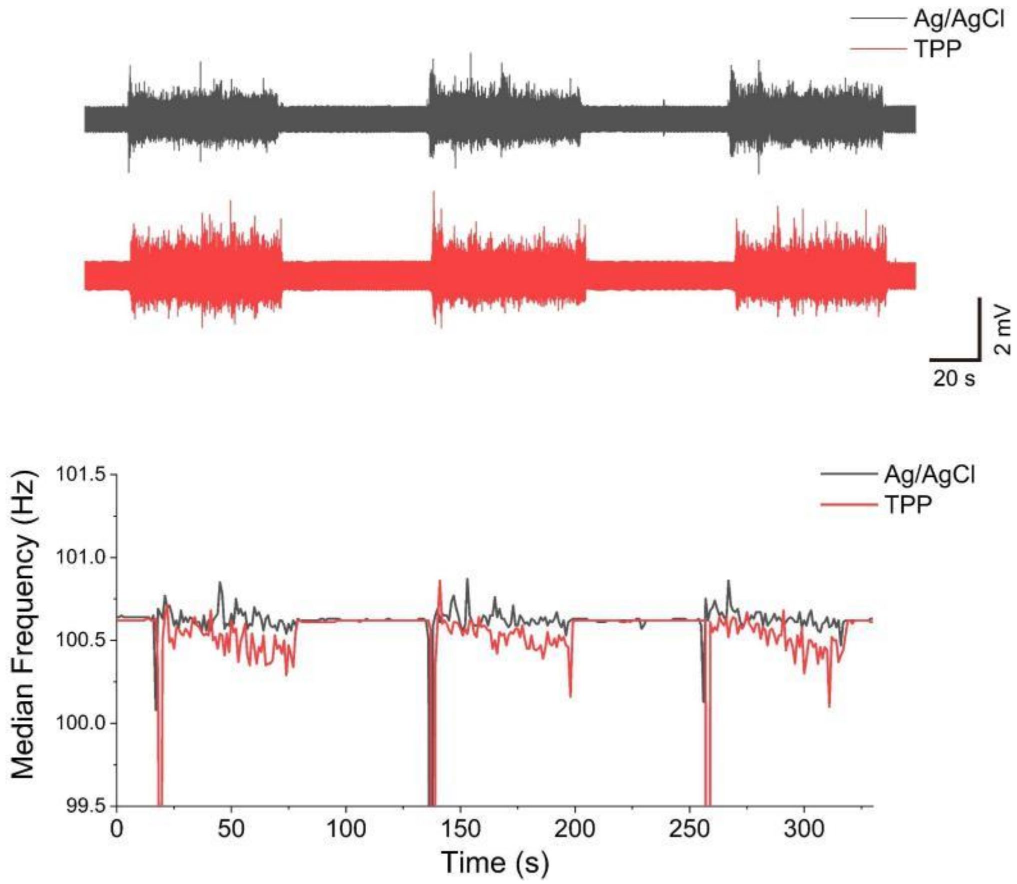
	Materials for electrodes	Materials for substrates	Is it intrinsically stretchable?	Strain	The adhesiveness of electrode (N/cm)	The number of channels	Success rate of channels	The smallest area of the electrode (mm^2)	Electrode-skin impedance at 100 Hz ($K\Omega*cm^2$)	Long-term test (Hour	RMS of Noise (μV)	Signal-to- noise ratio (dB)	Reference
	Ag/AgCl ink	Polypropylene	No	N/A	0	16	N/A	5.1	5000.0	N/A	N/A	~24.0	26
	Ag/AgCl ink	Polyethylene terephthalate	No	N/A	0	64	100%	14.5	N/A	2	N/A	~20.0	27
	Ag flakes/PDMS	PDMS	Yes	30%	0	8	100%	26.4	33.0	N/A	N/A	29.5	28
	Ag nanowires	PDMS	Yes	50%	0	18	94.4%	9.6	N/A	N/A	N/A	N/A	29
	Ag nanowires	Thermoplastic polyurethane	Yes	600%	0	4	100%	201.0	1004.8	N/A	~34.0	26.6	30
	Al	Polyethylene terephthalate	No	51%	0	16	100%	84.0	84.0	N/A	~130.0	N/A	31
	Au	Polyimide	No	40%	0	20	N/A	N/A	N/A	N/A	~300.0	~20.0	32
	Au	Polyimide	No	37%	0	64	N/A	0.8	117.0	N/A	~10.0	40.0±8.0	33
	Au	Polyimide	No	N/A	0	64	N/A	3.1	N/A	N/A	N/A	26.0±6.0	33
	Carbon/Silicone rubber	Textile	N/A	N/A	0	14	100%	400.0	320.0	N/A	~100.0	12.8±0.9	34
	MXene	PDMS	No	N/A	0	40	100%	7.1	2.4	N/A	~34.0	N/A	35
	MXene	Parylene-C	No	N/A	0	16	81.25%	2.6	256.0	N/A	~118.0	24.4±1.7	36
	PEDOT:PSS/Cholinium lactate	Kapton	No	N/A	0	16	100%	2.6	45.0	N/A	~40.0	15.6	37
	Stainless steel	Textile	No	N/A	0	150	90.6%	113.0	N/A	N/A	50.6±14.8	30.8±2.4	38
This work	**PEDOT/Polyvinly achohol/Tannic acid/Liquid metal**	**PDMS**	**Yes**	**188%**	**0.58**	**≥24**	**100%**	**0.8**	**80**	**120**	**1.0**	**42.3±0.7**	

One of these works was released in 2020, nine were released in 2021, and three were released in 2022. We can see from this table that the only electrode having adhesiveness is ours. Also, the patch’s 188% strain is higher than that of the majority of prior works, further assuring conformal contact with the skin. For the lowest noise RMS for sEMG recording by MEAP in this comparison, both adhesiveness and stretchability are crucial. The straightforward construction and reliable operation of MEAP allow it to be employed for a variety of muscles and situations in addition to enhanced signal capture. Taking into account the foregoing discussion, we believe our work to be state-of-the-art in the sEMG array sector.

This leaves the property of stretchability, which according to the authors should improve the quality of sEMG signals. The advantages and disadvantages of *stretchable electrode arrays have been debated among sEMG experts for many years. The problem is, that the interelectrode distance changes when the array is stretched. This affects the frequency spectrum of the sEMG signal in the case of a bipolar lead. Investigations in the frequency domain, as suggested by the authors for fatigue detection, are therefore not valid for non-isometric contractions, as it is not possible to exclude beyond doubt that a measured change in the frequency domain is not due to a change in the electrode distance. Stretchable electrode arrays are therefore fundamentally unsuitable for such applications.*

Our response:

Yes, we concur with the reviewer's worry concerning the impact of IED changes on sEMG signals. In bipolar mode of recording, it is true that changing the IED between two electrodes can produce RMS and frequency changes. But the recording setups are monopolar for most applications employing MEAP in this manuscript, including identifying muscle loading, exhaustion, and muscular activity map. Signals from each electrode are unrelated to one another. About the reviewer's concern concerning fatigue assessment for non-isometric contractions, it is claimed that a reduction in median frequency is also noticed via monopolar recording (Ref. 39). This was also seen in our trials (see Fig. 8g). The variation in slope values under varied loads revealed the muscle's various exhaustion stages. Extra figure below shows monopolar recording data from the new fatigue trial described in Fig. 4e. Even though the signals were heavily influenced by powerline noise and harmonics, the median frequencies reduced during the contraction. All of these references and demonstrations lead us to conclude that our stretchable electrode array is not only appropriate, but also offers an unrivaled advantage for such applications due to its superior attachment performance (see Supplementary Video 2 and 3).



Extra figure for reviewer 3's comment. Red dot 2223 3M electrodes and TPP electrodes were placed to identical positions on the FCU, and monopolar recording mode was chosen, and the median frequency was calculated. Both electrodes recorded decline in median frequency.

This brings me to another problem concerning the manuscript. The manuscript is full of claims - often in the superlative - about signal quality and possible applications of the described electrode, which are not statistically proven. They seem to be the purely subjective perceptions of the authors. This becomes particularly clear in Fig. 2 k, in which a signal with a motion artefact, which occur from time to time but not regularly, was compared with the signal detected with the introduced electrode. The manuscript does not describe whether and if so how repeat measurements were carried out and how these were statistically evaluated to substantiate the statements made.

Our response:

We thank the reviewer asking for clarification. The goal of Fig. 2k is to demonstrate that adhesive TPP electrodes are superior to our non-adhesive PEDOT-PVA electrode for sEMG recording. We appreciate the reviewer's worry that motion artifacts occur irregularly due to the wire-swinging effect. Nevertheless, the motion artefact in this situation is mostly created by the relative movement of electrodes on the skin during muscle movements. As can be seen in the Supplementary Fig. 8, the motion artifacts are most severe during the biceps curl, whether recorded in monopolar or bipolar mode. We also discovered that sticky TPP electrodes produce less motion artifacts than non-adhesive PEDOT-PVA electrodes. We used statistical tools to examine 5 curls for each type of electrode and discovered that the RMS change differed between them, proving our claim: the adhesive TPP electrodes are more suitable for sEMG recording than our non-adhesive PEDOT-PVA electrode. About the reviewer's worry about subjective conclusions, we accept that the lack of specifics in the experimental portion contributed to this perception to some extent. Nonetheless, we need to point out that all our findings are based on the outcomes of tests or comparative studies with other publications. The SNR values in Fig. 2l were obtained from various sources and compared. In Fig. 3d, three contractions were recorded every day to provide statistical RMS and SNR values. In Fig. 8f, all RMS values collected throughout the isometric exercise were used to create a box chart. In addition, we tested three individuals in total to assess our MEAP performance on various persons for injury prevention. We always strive to make things objective by using statistical analysis of our findings.

We included information in ‘sEMG signal recording’ for better explanations to make things clearer.

See page 34 ‘During the biceps brachii muscle recordings, a comparative test between PEDOT-PVA and TPP electrodes was performed. The subject was instructed to keep the curl speed between flexion and extension at 4.5 rad/s. Five contractions of each electrode were tested for RMS alterations. For long-term test, three contractions were recorded each day to provide statistical RMS and SNR values.’

Fig. 4 a and e shows another problem that arises when characterising the quality of the novel electrodes. The electrodes of the devices used for comparison are not located in the same position as the novel electrodes. Rather, the comparison signals are derived at less favourable positions, which has a negative influence on the signal *amplitude, the SNR and the frequency spectrum. An objective comparison between the two devices is not possible under these conditions.*

Our response: We appreciate the reviewer's thoughtful comment on the experiment design of this comparison test. We were also worried before that whether such position difference caused difference in signal recording. The reason we chose to record contractions by two types of electrodes simultaneously is we hope to get 'identical' signals and do analysis on the fatigue which should only be influenced by the type of electrodes. We were more worried that each contraction itself would have a difference in fatigue which introduced another variable into the analysis even if the tasks were the same. So here is the issue of choice. We assumed remarkably close positions of two electrodes would not make too much difference to the signal recorded, so we chose that way originally. But we also agree with the reviewer's opinion, so we conducted this experiment again to eradicate the position effect, trying to make an objective comparison as the reviewer wished. We need to clarify that through this new test, we cannot get the same conclusion that TPP electrodes are better for fatigue measurements than Ag/AgCl electrodes, but we can say the TPP electrodes can measure the muscle fatigue just as Ag/AgCl electrodes. It is worth mentioning that we changed the Ag/AgCl electrodes from Foam Monitoring 2228 to Red Dot 2223 electrodes in the new experiments. We updated Fig. 4 with the new experiment.

See page 14: 'To assess the ability of TPP electrodes to obtain information of frequency in the signal, Ag/AgCl and TPP electrodes were set on the same position on FCU, and the subject was asked to curl the wrist with a 5 kg dumbbell for three long periods to activate FCU (Fig. 4e). The TPP electrodes showed a little better SNR than the Ag/AgCl electrodes that they are 39.2, 37.5 and 40.5 dB for three contractions recorded by TPP electrodes and 38.9, 37.5 and 38.6 dB by Ag/AgCl electrodes. The spectrograms showed TPP electrodes can give clear frequency information just like Ag/AgCl electrodes (Fig. 4f). To compare the performances of Ag/AgCl and TPP electrodes on fatigue measurement, the subject was asked to curl the wrist 60 s for three times for each type of electrodes. Three tasks were named as flexion 1, 2 and 3 to calculate median frequency during each task (Fig. 4g). Linear fittings were made for first 25 s of each contraction, to quantify the outcome with less errors[73]. The slopes obtained by two types of electrodes both showed negative which indicated the muscle was in fatigue. This test proved the TPP electrodes can measure the muscle fatigue the same as Ag/AgCl electrodes.’

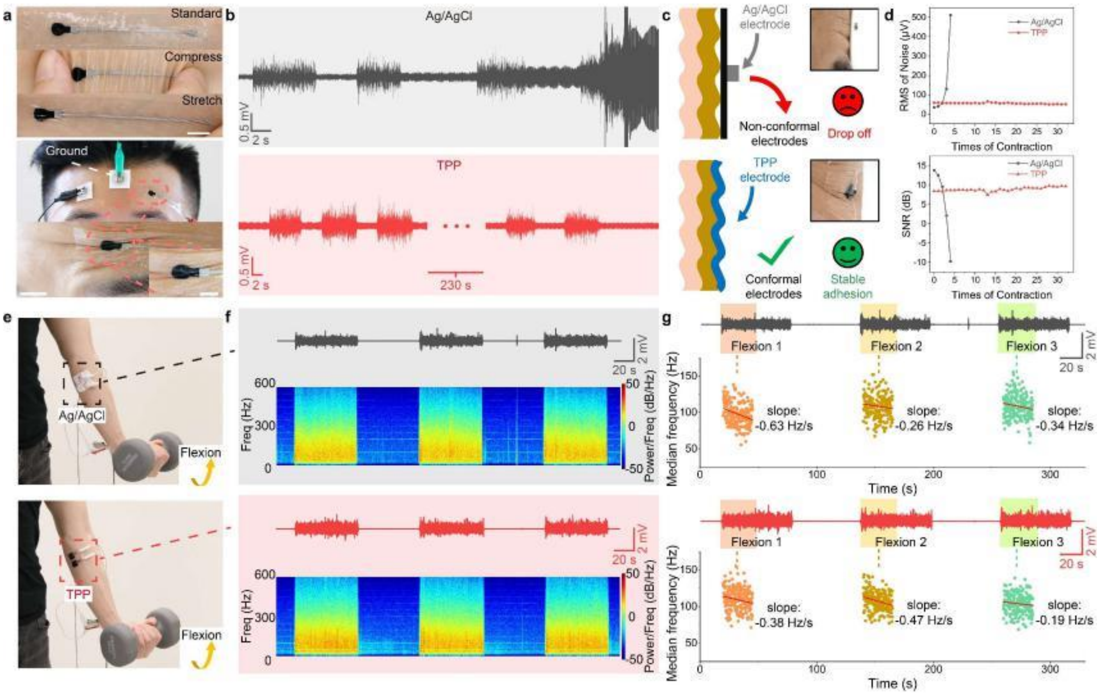


Fig. 4 Comparison of recording performances on skin between Ag/AgCl and TPP electrodes.

a Up, standard, compressing and stretching TPP electrodes on the skin. Scale bar: 1 cm; bottom, photographs of Ag/AgCl and TPP electrodes when recording sEMG of frontalis and the TPP electrode in the skin folds. Scale bar of photo at the bottom: 1 cm; bottom inset: 0.5 cm.

b sEMG signals recorded by Ag/AgCl and TPP electrodes, respectively. The subject was asked to make each contraction for 5 seconds. In the case of recording by Ag/AgCl electrodes, after four times of contraction, noises were even higher than signals.

c Schematic illustrations and lateral photos of Ag/AgCl electrode and TPP electrode on skin folds.

d Noise level and SNR recorded by two electrodes during contractions.

e Photographs of electrode configuration on FCU and contraction task. Two pairs of electrodes were attached the same position on the forearm.

f sEMG signals and spectrograms recorded by Ag/AgCl and TPP electrodes respectively.

g sEMG signals and fitting results of median frequency during flexion 1, 2 and 3 recorded by Ag/AgCl and TPP electrodes. Decreasing median frequencies indicated fatigue of the muscle.

Finally, a comment on electrode arrays. The use of electrode arrays has been known for a long time under the pseudonym High Density sEMG (HDsEMG) and is widely used in different research questions. The method is called sEMG imaging and the "heat maps" shown in Fig. 5 g and h are called "muscle activity maps" in the literature. The fact that HDsEMG is suitable for localising anatomical structures such as the neuromuscular junction or tendon insertion has been known since the 1990s and has been studied in a number of different investigations. In connection with fatigue and pain, a change in the spatial distribution of the activity of the muscle has already been demonstrated, as well as a change in the spectrum of the signal. This fundamental work is not mentioned anywhere in the manuscript. Rather, the impression is given that such investigations are only made possible by the new type of electrode.

Our response: We appreciate the reviewer's generous sharing about the knowledge of HDsEMG. Yes, we agree the reviewer's point that HDsEMG is already widely used in different research questions. The objective of showing our results in original Fig. 5 is also to prove MEAP can complete the task carried before by HDsEMG, just like we stated in the paragraph: 'Such tools will create remarkable benefits and provide a new means for clinical diagnosis, medical treatment and sports sciences.' The reason we said this is our stretchable array indeed can do those investigations which our new type of electrode can complete better, but much harder by traditional HDsEMG. We showed them in the new Supplementary Video 3 that traditional HDsEMG is extremely easy to fall off when it is attached on the muscle-tendon junction. So using traditional HDsEMG makes less stable recording for sEMG and let alone the monitoring of tendon displacement under the skin.

As for the reviewer's concern on 'HDsEMG is suitable for localising anatomical structures such as the neuromuscular junction or tendon insertion', we searched on Pubmed (https://pubmed.ncbi.nlm.nih.gov/) with key words 'EMG', 'array' and 'tendon', there are only 33 results and none of them studied on the location of muscle-tendon junction (Ref. 40-72), which is also different to two terms the reviewer proposed. Based on that, we believe this application by MEAP is novel. We also used it to help our injury prevention analysis which should prove that our stretchable electrode array has potential for future applications in many areas such as clinical diagnosis, medical treatment and sports sciences.

We changed ‘heatmaps’ to ‘muscle activity maps’ in manuscript.

See page 19: ‘Muscle activity maps based on RMS values were generated to visualize the advantage of high-density systems.’

‘... which caused the active zone to move to the right in the muscle activity maps with an increased activity.’

page 21: ‘**g, h** Muscle activity maps of sEMG recorded ...’

Page 34: ‘Muscle contraction task for muscle activity maps’, ‘Muscle activity maps were generated to help visualize the change in activity during the task’.

For fundamental work about fatigue and pain, we added simple introduction in the manuscript because this is not the focus of this manuscript.

See page 2: ‘There are also many works studied on neuromuscular junctions by high density sEMG, to demonstrate the muscle fatigue and pain[17–21].’

17. Muceli, S. & Farina, D. Simultaneous and proportional estimation of hand kinematics from EMG during mirrored movements at multiple degrees-of-freedom. *IEEE Trans. Neural Syst. Rehabil. Eng.* **20**, 371–378 (2012).

18. Muceli, S., Falla, D. & Farina, D. Reorganization of muscle synergies during multidirectional reaching in the horizontal plane with experimental muscle pain. *J. Neurophysiol.* **111**, 1615–1630 (2014).

19. Merletti, R., Rainoldi, A. & Farina, D. Surface electromyography for noninvasive characterization of muscle. *Exerc. Sport Sci. Rev.* **29**, 20–25 (2001).

20. Merletti, R., Farina, D. & Gazzoni, M. The linear electrode array: A useful tool with many applications. *J. Electromyogr. Kinesiol.* **13**, 37–47 (2003).

21. Merletti, R. *et al.* Multichannel surface EMG for the non-invasive assessment of the anal sphincter muscle. *Digestion* **69**, 112–122 (2004).

We thank reviewer’s incisive and thoughtful comments again for pointing out the issues in the manuscript. We feel that our manuscript is more convincible, and the advantages of MEAP are more unrivaled after we supplemented two experiments according to the reviewer’s suggestions.

Reference for the response to the Reviewer #3

1. Harati, A. & Jahanshahi, A. A reliable stretchable dry electrode for monitoring of EEG signals. *Sensors Actuators A Phys.* **326**, 112727 (2021).

2. Alban, M. V., Lee, H., Moon, H. & Yoo, S. Micromolding fabrication of biocompatible dry micro-pyramid array electrodes for wearable biopotential monitoring. *Flex. Print. Electron.* **6**, 045008 (2021).

3. Jiang, Y. *et al.* Flexible and stretchable dry active electrodes with pdms and silver flakes for bio-potentials sensing systems. *IEEE Sens. J.* **21**, 12255–12268 (2021).

4. Yoon, S. *et al.* Highly stretchable metal-polymer hybrid conductors for wearable and self-cleaning sensors. *NPG Asia Mater.* **13**, 4 (2021).

5. Yun, I. *et al.* Stable Bioelectric Signal Acquisition Using an Enlarged Surface-Area Flexible Skin Electrode. *ACS Appl. Electron. Mater.* **3**, 1842–1851 (2021).

6. Li, Q. *et al.* Highly Thermal-Wet Comfortable and Conformal Silk-Based Electrodes for On-Skin Sensors with Sweat Tolerance. *ACS Nano* **15**, 9955–9966 (2021).

7. Tang, W. *et al.* Delamination-Resistant Imperceptible Bioelectrode for Robust Electrophysiological Signals Monitoring. *ACS Mater. Lett.* **3**, 1385–1393 (2021).

8. Won, Y. *et al.* Biocompatible, Transparent, and High-Areal-Coverage Kirigami PEDOT:PSS Electrodes for Electrooculography-Derived Human-Machine Interactions. *ACS Sensors* **6**, 967–975 (2021).

9. Blau, R. *et al.* Intrinsically Stretchable Block Copolymer Based on PEDOT:PSS for Improved Performance in Bioelectronic Applications. *ACS Appl. Mater. Interfaces* **14**, 4823–4835 (2022).

10. Zhou, X. *et al.* Self-healing, stretchable, and highly adhesive hydrogels for epidermal patch electrodes. *Acta Biomater.* **139**, 296–306 (2022).

11. Wang, S. *et al.* Self-adhesive, stretchable, biocompatible, and conductive nonvolatile eutectogels as wearable conformal strain and pressure sensors and biopotential electrodes for precise health monitoring. *ACS Appl. Mater. Interfaces* **13**, 20735–20745 (2021).

12. Zhang, L. *et al.* Fully organic compliant dry electrodes self-adhesive to skin for long-term motion-robust epidermal biopotential monitoring. *Nat. Commun.* **11**, 4683 (2020).

13. Cao, J. *et al.* Stretchable and Self-Adhesive PEDOT:PSS Blend with High Sweat Tolerance as Conformal Biopotential Dry Electrodes. *ACS Appl. Mater. Interfaces* **14**, 39159–39171 (2022).

14. Roldan-Vasco, S., Restrepo-Agudelo, S., Valencia-Martinez, Y. & Orozco-Duque, A. Automatic detection of oral and pharyngeal phases in swallowing using classification algorithms and multichannel EMG. J. Electromyogr. Kinesiol. 43, 193–200 (2018).

15. Zhuang, M. et al. Highly Robust and Wearable Facial Expression Recognition via Deep-Learning-Assisted, Soft Epidermal Electronics. Research 2021, (2021).

16. Fortune, B. C., Pretty, C. G., Chatfield, L. T., McKenzie, L. R. & Hayes, M. P. Data captured using low-cost active electromyography. Data Br. 29, 105239 (2020).

17. Becerra-Fajardo, L. et al. Floating EMG sensors and stimulators wirelessly powered and operated by volume conduction for networked neuroprosthetics. J. Neuroeng. Rehabil. 19, 57 (2022).

18. Ye-Lin, Y. et al. Directed Functional Coordination Analysis of Swallowing Muscles in Healthy and Dysphagic Subjects by Surface Electromyography. Sensors 22, (2022).

19. Frenzel, D., Greim, C. A., Sommer, C., Bauerle, K. & Roewer, N. Is the bispectral index appropriate for monitoring the sedation level of mechanically ventilated surgical ICU patients? Intensive Care Med. 28, 178–183 (2002).

20. Campos, D. P. et al. Short-term fibre intake estimation in goats using surface electromyography of the masseter muscle. Biosyst. Eng. 183, 209–220 (2019).

21. Phinyomark, A., Phukpattaranont, P. & Limsakul, C. Fractal analysis features for weak and single-channel upper-limb EMG signals. Expert Syst. Appl. 39, 11156–11163 (2012).

22. Wernbom, M., Järrebring, R., Andreasson, M. A. & Augustsson, J. Acute Effects of Blood Flow Restriction on Muscle Activity and Endurance During Fatiguing Dynamic Knee Extensions at Low Load. J. Strength Cond. Res. 23, 2389–2395 (2009).

23. Allen, D. P. A frequency domain Hampel filter for blind rejection of sinusoidal interference from electromyograms. J. Neurosci. Methods 177, 303–310 (2009).

24. Lee, S. et al. Wireless Epidermal Electromyogram Sensing System. Electronics 9, 269 (2020).

25. Shim, J. K., Choi, H. S. & Shin, J. H. Effects of neuromuscular training on knee joint stability after anterior cruciate ligament reconstruction. J. Phys. Ther. Sci. 27, 3613–3617 (2015).

26. Cantu, E. *et al.* Printed Multi-EMG Electrodes on the 3D Surface of an Orthosis for Rehabilitation: A Feasibility Study. *IEEE Sens. J.* 21, 14407–14417 (2021).

27. Moin, A. *et al.* A wearable biosensing system with in-sensor adaptive machine learning for hand gesture recognition. *Nat. Electron.* 4, 54–63 (2021).

28. Zhu, K., Guo, W., Yang, G., Li, Z. & Wu, H. High-Fidelity Recording of EMG Signals by Multichannel On-Skin Electrode Arrays from Target Muscles for Effective Human-Machine Interfaces. *ACS Appl. Electron. Mater.* 3, 1350–1358 (2021).

29. Lin, Y. *et al.* High-resolution and large-size stretchable electrodes based on patterned silver nanowires composites. *Nano Res.* 15, 4590–4598 (2022).

30. Zhao, H. *et al.* Ultrastretchable and Washable Conductive Microtextiles by Coassembly of Silver

Nanowires and Elastomeric Microfibers for Epidermal Human-Machine Interfaces. *ACS Mater. Lett.* 3, 912–920 (2021).

31. Gandla, S. *et al.* Ultrafast Prototyping of Large-Area Stretchable Electronic Systems by Laser Ablation Technique for Controllable Robotic Arm Operations. *IEEE Trans. Ind. Electron.* 69, 4245–4253 (2022).

32. Mei, Z., Zhao, N., Yang, B. & Liu, J. Flexible Concentric Ring Electrode Array for Low-Noise and Non-Invasive Detection. in *Proceedings of the IEEE International Conference on Micro Electro Mechanical Systems (MEMS)* vols 2021-January 266–269 (Institute of Electrical and Electronics Engineers Inc., 2021).

33. Chandra, S. *et al.* Performance Evaluation of a Wearable Tattoo Electrode Suitable for High-Resolution Surface Electromyogram Recording. *IEEE Trans. Biomed. Eng.* 68, 1389–1398 (2021).

34. Alizadeh-Meghrazi, M. *et al.* A Mass-Producible Washable Smart Garment with Embedded Textile EMG Electrodes for Control of Myoelectric Prostheses: A Pilot Study. *Sensors* 22, 666 (2022).

35. Driscoll, N. *et al.* MXene-infused bioelectronic interfaces for multiscale electrophysiology and stimulation. *Sci. Transl. Med.* 13, eabf8629 (2021).

36. Murphy, B. B. *et al.* A Gel-Free Ti3C2Tx-Based Electrode Array for High-Density, High-Resolution Surface Electromyography. *Adv. Mater. Technol.* 5, 2000325 (2020).

37. Velasco-Bosom, S. *et al.* Conducting Polymer-Ionic Liquid Electrode Arrays for High-Density Surface Electromyography. *Adv. Healthc. Mater.* 10, 2100374 (2021).

38. Ting, J. E. *et al.* Sensing and decoding the neural drive to paralyzed muscles during attempted movements of a person with tetraplegia using a sleeve array. *J. Neurophysiol.* 127, 2104–2118 (2021).

39. Kleine, B. U., Schumann, N. P., Stegeman, D. F. & Scholle, H. C. Surface EMG mapping of the human trapezius muscle: The topography of monopolar and bipolar surface EMG amplitude and spectrum parameters at varied forces and in fatigue. Clin. Neurophysiol. 111, 686–693 (2000).

40. Carraro, U. Thirty years of translational research in Mobility Medicine: Collection of abstracts of the 2020 Padua Muscle Days. Eur. J. Transl. Myol. 30, 8826 (2020).

41. Huang, C., Chen, X., Cao, S. & Zhang, X. Muscle-tendon units localization and activation level analysis based on high-density surface EMG array and NMF algorithm. J. Neural Eng. 13, 66001 (2016).

42. Chen, X., Wang, S., Huang, C., Cao, S. & Zhang, X. ICA-based muscle-tendon units localization

and activation analysis during dynamic motion tasks. Med. Biol. Eng. Comput. 56, 341–353 (2018).

43. Zhang, X. et al. EMG-Torque Relation in Chronic Stroke: A Novel EMG Complexity Representation With a Linear Electrode Array. IEEE J. Biomed. Heal. informatics 21, 1562–1572 (2017).

44. Thompson, C. K., Johnson, M. D., Negro, F., Farina, D. & Heckman, C. J. Motor Unit Discharge Patterns in Response to Focal Tendon Vibration of the Lower Limb in Cats and Humans. Front. Integr. Neurosci. 16, 836757 (2022).

45. McAvoy, M. et al. Flexible Multielectrode Array for Skeletal Muscle Conditioning, Acetylcholine Receptor Stabilization and Epimysial Recording After Critical Peripheral Nerve Injury. Theranostics 9, 7099–7107 (2019).

46. Li, W. & Sakamoto, K. The influence of location of electrode on muscle fiber conduction velocity and EMG power spectrum during voluntary isometric contraction measured with surface array electrodes. Appl. Human Sci. 15, 25–32 (1996).

47. Merletti, R., Farina, D. & Gazzoni, M. The linear electrode array: a useful tool with many applications. J. Electromyogr. Kinesiol. Off. J. Int. Soc. Electrophysiol. Kinesiol. 13, 37–47 (2003).

48. English, A. W., Wolf, S. L. & Segal, R. L. Compartmentalization of muscles and their motor nuclei: the partitioning hypothesis. Phys. Ther. 73, 857–867 (1993).

49. Wakeling, J. M. The recruitment of different compartments within a muscle depends on the mechanics of the movement. Biol. Lett. 5, 30–34 (2009).

50. Tokunaga, T. [Muscle fiber conduction velocity and frequency parameters of surface EMG during fatigue of the human masseter muscle. 2. Frequency parameters]. Nihon Hotetsu Shika Gakkai Zasshi 33, 804–817 (1989).

51. Farina, D. & Merletti, R. A novel approach for estimating muscle fiber conduction velocity by spatial and temporal filtering of surface EMG signals. IEEE Trans. Biomed. Eng. 50, 1340–1351 (2003).

52. Thompson, C. K. et al. Robust and accurate decoding of motoneuron behaviour and prediction of the resulting force output. J. Physiol. 596, 2643–2659 (2018).

53. Botter, A., Vieira, T. M., Geri, T. & Roatta, S. The peripheral origin of tap-induced muscle contraction revealed by multi-electrode surface electromyography in human vastus medialis. Sci. Rep. 10, 2256 (2020).

54. Merletti, R., Roy, S. H., Kupa, E., Roatta, S. & Granata, A. Modeling of surface myoelectric signals--Part II: Model-based signal interpretation. IEEE Trans. Biomed. Eng. 46, 821–829 (1999).

55. Ye, X., Beck, T. W. & Wages, N. P. Prolonged passive static stretching-induced innervation zone shift in biceps brachii. Appl. Physiol. Nutr. Metab. = Physiol. Appl. Nutr. Metab. 40, 482–488 (2015).

56. De la Fuente, C. et al. Distal overactivation of gastrocnemius medialis in persistent plantarflexion weakness following Achilles tendon repair. J. Biomech. 148, 111459 (2023).

57. Minetto, M. A., Botter, A., Ravenni, R., Merletti, R. & De Grandis, D. Reliability of a novel neurostimulation method to study involuntary muscle phenomena. Muscle Nerve 37, 90–100 (2008).

58. Bujalski, P., Martins, J. & Stirling, L. A Monte Carlo analysis of muscle force estimation sensitivity to muscle-tendon properties using a Hill-based muscle model. J. Biomech. 79, 67–77 (2018).

59. Barandun, M., von Tscharner, V., Meuli-Simmen, C., Bowen, V. & Valderrabano, V. Frequency and conduction velocity analysis of the abductor pollicis brevis muscle during early fatigue. J. Electromyogr. Kinesiol. Off. J. Int. Soc. Electrophysiol. Kinesiol. 19, 65–74 (2009).

60. Mito, K. & Sakamoto, K. On the evaluation of muscle fiber conduction velocity considering waveform properties of an electromyogram in M. biceps brachii during voluntary isometric contraction. Electromyogr. Clin. Neurophysiol. 42, 137–149 (2002).

61. Mananas, M. A., Rojas, M., Mandrile, F. & Chaler, J. Evaluation of muscle activity and fatigue

1002 in extensor forearm muscles during isometric contractions. Conf. Proc. ... Annu. Int. Conf.
IEEE

Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf. 2005, 5824–5827 (2005).

62. Sakamoto, K. & Li, W. Effect of muscle length on distribution of muscle fiber conduction velocity for M. biceps brachii. Appl. Human Sci. 16, 1–7 (1997).

63. Farina, D., Arendt-Nielsen, L., Merletti, R. & Graven-Nielsen, T. Assessment of single motor unit conduction velocity during sustained contractions of the tibialis anterior muscle with advanced spike triggered averaging. J. Neurosci. Methods 115, 1–12 (2002).

64. Rodriguez-Falces, J., Botter, A., Vieira, T. & Place, N. The M waves of the biceps brachii have a stationary (shoulder-like) component in the first phase: implications and recommendations for M-wave analysis. Physiol. Meas. 42, 15007 (2021).

65. Rojas, M., Mañanas, M. A., Muller, B. & Chaler, J. Activation of forearm muscles for wrist extension in patients affected by lateral epicondylitis. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Int. Conf. 2007, 4858–4861 (2007).

66. Mito, K. & Sakamoto, K. Distribution of muscle fiber conduction velocity of m. masseter during voluntary isometric contraction. Electromyogr. Clin. Neurophysiol. 40, 275–285 (2000).

67. Cho, S.-G. et al. Hand motion recognition based on forearm deformation measured with a distance sensor array. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Int. Conf. 2016, 4955–4958 (2016).

68. Masuda, T. & Sadoyama, T. The propagation of single motor unit action potentials detected by a surface electrode array. Electroencephalogr. Clin. Neurophysiol. 63, 590–598 (1986).

69. Nishihara, K., Futami, T., Hosoda, K. & Gomi, T. Validation of estimated muscle fiber conduction velocity with the normalized peak-averaging technique. J. Electromyogr. Kinesiol. Off. J. Int. Soc. Electrophysiol. Kinesiol. 15, 93–101 (2005).

70. Li, W. & Sakamoto, K. Distribution of muscle fiber conduction velocity of M. biceps brachii during voluntary isometric contraction with use of surface array electrodes. Appl. Human Sci. 15, 41–53 (1996).

71. Sakamoto, K. & Mito, K. Muscle fiber conduction velocity during isometric contraction and the recovery period. Electromyogr. Clin. Neurophysiol. 40, 151–161 (2000).

72. Hník, P., Vejsada, R. & Macková, E. V. EMG activity in ‘compensatory’ muscle hypertrophy. Physiol. Bohemoslov. 35, 285–288 (1986).

REVIEWER COMMENTS

Reviewer #1 (Remarks to the Author):

The authors have made well addressed my concerns in the revised manuscript. The quality of the revised version of this study is worthy of publishing in Nature Communcations.

Reviewer #2 (Remarks to the Author):

Here are some other comments which should be further addressed before its consideration for publication:

1. Could be the MEAP be used repetitively? (I mean the 4*6 array, not the PPT electrode)

2. How about the permeability of the MEAP since the authors claimed its long-term usage?

3. For the reader's better understanding, more details are expected on the MEAP connected to the data acquisition module via flexible printable circuit board connectors.

4. The authors compared the attachment performance between the commercial array and MEAP. In fact, the spatial density and distribution of the electrode sites is also important to its application. The commercial array has 64 channels with 8*8 array while the MEAP is 24 with 4*6 array, which leads to mismatch location of the electrode sites during comparison. It would be better to make a comparison directly with the same configuration.

5. In addition to the innovative aspects of the PPT electrode and the MEAP, please also specify the corresponding limitations at Conclusion Section.

Reviewer #3 (Remarks to the Author):

The authors have tried to address some of my concerns. However, they only succeeded to a very limited extent. The manuscript credibly demonstrates that the mechanical properties of the new electrode array are superior to other electrode arrays. The same applies to the purely electrical properties. However, the manuscript still has major shortcomings when the suitability of the new electrode array is examined with regard to the detection of sEMG signals and compared with state-of-the-art methods. As an example, I refer again to Fig. 4e, which still compares the sEMG signals derived with the new electrodes with signals from sEMG derivations that do not comply with the international recommendations for the derivation of sEMG signals. The international recommendations for the detection of sEMG signals are defined and described in the SENIAM and CEDE projects. The project results are published as well as described on the internet. E.g. it seems, that the electrode distance is larger than 1/4 of the muscle fibre length of the FCU and that the sensor is placed halfway the (most) distal motor endplate zone and the distal tendon (SENIAM Recommendations). In addition, any information about the sensor used, such as size of the active area, interelectrode distance and exact position of the electrodes as well as the treatment of the skin, is missing. This information is essential to evaluate the quality of the derived sEMG signals and is standard in any sEMG publication, even when comparing two methods Additionally, open cable clips were used to fix the cables to the electrodes. This worsens the SNR considerably. Shielded connectors are more common.

The manuscript is still full of claims - often in the superlative - about signal quality and possible applications of the described electrode, which are not statistically proven. I would like to draw special attention to the statistics here once again! I do not see any statistical calculation in the entire manuscript that shows that the claims made with respect to application are statistically significant, or that at least a tendency can be read. Instead, individual examples are shown, which is good, but does not justify the strong formulations and the emphasis on the new electrode over existing methods. That the major problem with the paper is the lack of statistics. As it stands, there is no evidence for the results in the different applications, and for me that is a no-go for a publication.

Furthermore, some of the claims about the state of the art of existing sEMG electrode/lead methods are simply wrong. Let me cite the abstract: "However, current sEMG electrodes do not offer adequate high-quality data for their widespread use in clinics and everyday life, since these are neither stretchable nor arrayed". This sentence is simply wrong! Electrode arrays have been available for many years and some of them are stretchable. Whether stretchability improves the clinical significance of the sEMG is unknown, while the clinical benefit of the array is proven. The manuscript is full of such examples and should be formulated in a less absolute way to reflect reality.

Response to reviewers for the manuscript (NCOMMS-22-46103B-Z)

Reviewer #1 (Remarks to the Author):

The authors have made well addressed my concerns in the revised manuscript. The quality of the revised version of this study is worthy of publishing in Nature Communications.

Our response: We appreciate the reviewer taking the time to carefully read our revised manuscript and provide such excellent feedback.

Reviewer #2 (Remarks to the Author):

Here are some other comments which should be further addressed before its consideration for publication:

Our response: We really value the reviewer's time spent reading our revised manuscript thoroughly and providing other comments to further improve the manuscript.

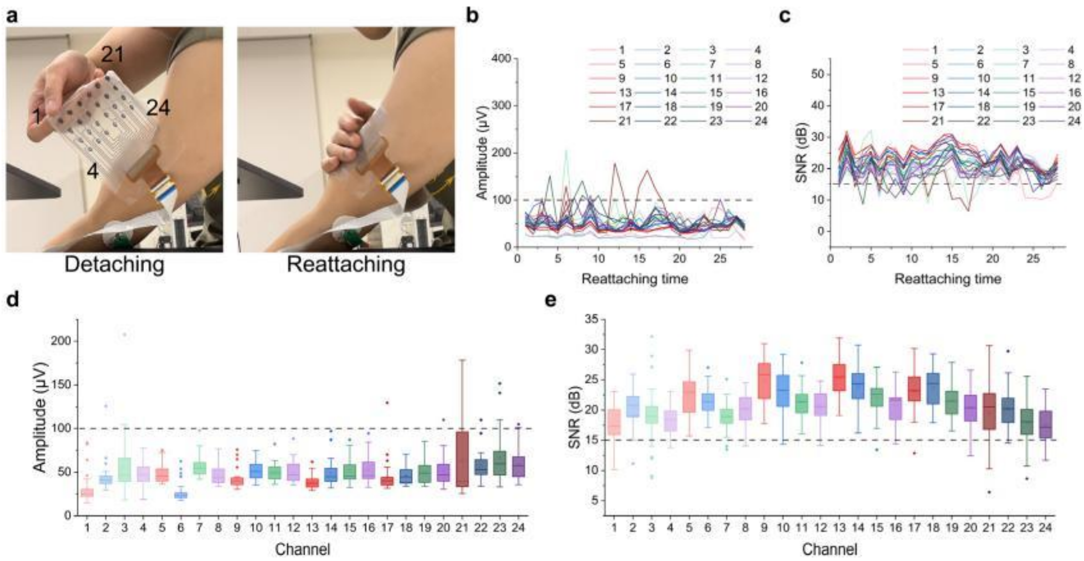
*1. Could be the MEAP be used repetitively? (I mean the 4*6 array, not the PPT* *electrode)*

Our response: We appreciate the reviewer seeking further clarification. In the previous response letter, we proved TPP electrodes can be used repetitively by different tests. We have also referenced our earlier publication[1], which demonstrates the reattachment of the substrate part on the skin repetitively. By combining these findings, we aimed to convey that the MEAP is suitable for repetitive use.

To provide a more convincing answer to the reviewer's question, we have designed a new experiment specifically to demonstrate the repetitive use of the MEAP on the skin. However, it should be noted that the signal qualities obtained from this experiment were slightly lower compared to previous tests conducted using TPP electrodes. This is primarily due to the much smaller contact area of the channels in the MEAP. Nevertheless, the signal qualities obtained were still sufficiently high (approximately 20 dB) for use after repetitive attachment.

While we acknowledge the value of the experiment suggested by the reviewer, we have decided that incorporating these results into the main body of the text would disrupt the current outline, as the manuscript primarily focuses on electrodes and the MEAP separately. Therefore, we have chosen to include these additional findings in the Supplementary Information section.

Our changes on Page 11: As for the MEAP, we checked the reattachment performance of the patch (Supplementary Fig. 14). The results showed all channels of MEAP have stable performances. This indicates that MEAP can be used repetitively.



Supplementary Fig. 14 The reattachment test of 24-channel MEAP on the skin.

a Images showing the process of detachment and reattachment of MEAP on biceps brachii.

b, c The baseline noise of each channel and the SNR of each channel plotted for each reattachment. The baseline noise consistently maintained an amplitude of approximately 50 μV across all channels, even after 28 reattachments. Similarly, the SNR remained stable at 20 dB across all channels after 28 reattachments.

d, e Whisker plots of statistical verification of **b** and **c** for the 28 reattachments. Statistical analysis was conducted to assess the baseline noise and SNR of each channel throughout the reattachment test. 28 reattachments, per channel, were included in the analysis. The box plots depict the mean (center square), median (center line), 25th to

75th percentiles (box), and the lower and upper whiskers representing the smallest and largest values that are ≤ 1.5 times the interquartile range, respectively. Outliers are also shown.

2. How about the permeability of the MEAP since the authors claimed its long-term usage?

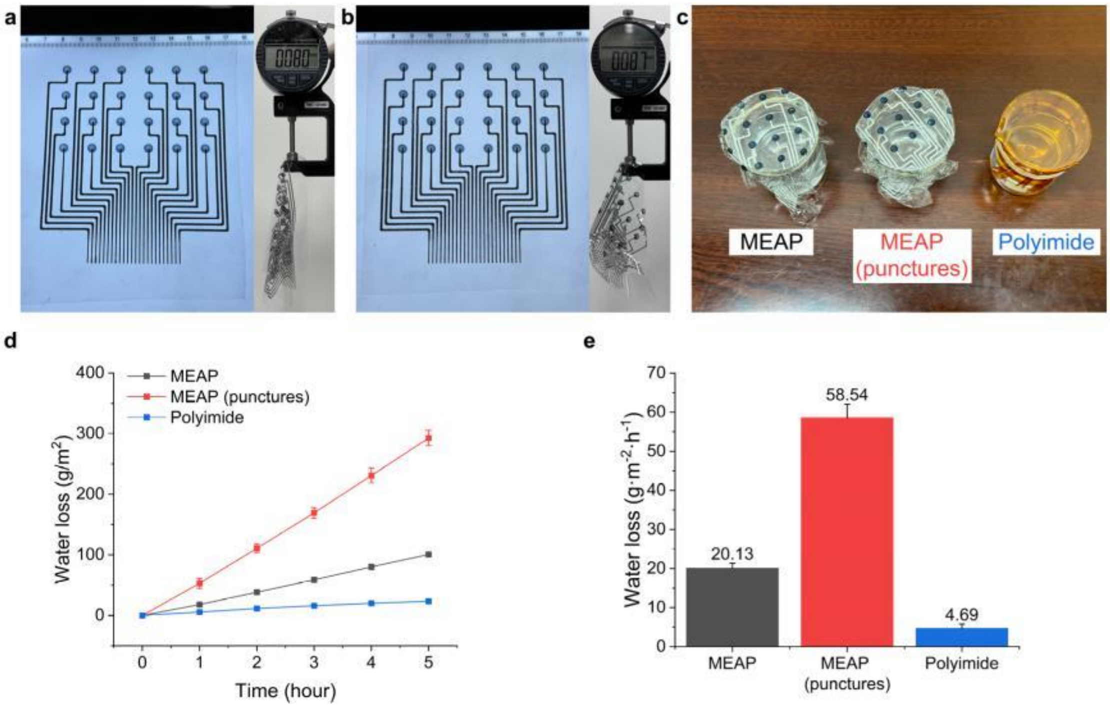
Our response: We appreciate the reviewer for highlighting the concern regarding permeability. The insensible sweat rate of individuals typically ranges from 12 to 42

$g \cdot m^{-2} \cdot h^{-1}$ (Ref. 2), indicating that materials with similar permeability can meet the requirements for daily use or long-term wearing. The substrate material used for the MEAP is PDMS, which inherently possesses permeability. Our results demonstrate that the MEAP has a permeability of approximately 20 $g \cdot m^{-2} \cdot h^{-1}$, which is suitable for the normal evaporation of sweat.

Moreover, if necessary, punctures can be made on the substrate to further enhance the permeability of the MEAP. This adjustment can be customized based on individual experiments and subject requirements. However, it is important to note that in comparison to PDMS, polyimide exhibited significantly lower permeability, implying limitations for long-term use.

As per the structure of our manuscript, similar to the first additional test, we have included these results in the Supplementary Information section. This decision allows us to maintain the coherence and flow of the main body text.

Our modifications on Page 11: We also examined the permeability performance of MEAP for daily long-term use (Supplementary Fig. 15). The results demonstrate that the permeability of the MEAP is well-suited for extended periods of usage, as it does not hinder the normal evaporation of sweat from the skin. Furthermore, we discovered that the permeability of the MEAP can be adjusted by modifying the physical structure of the substrate. This ability to tune the permeability enables us to create a comfortable wearing experience for daily use, as the permeability can be increased to a level that promotes adequate airflow.



Supplementary Fig. 15 The permeability comparison test between MEAP, MEAP (punctures) and polyimide.

a, b Images showing the MEAP and MEAP (punctures) with thicknesses of 80 and 87 μm, respectively. The MEAP (punctures) features 24 punctures (1 mm in diameter), corresponding to the number of TPP electrodes on the patch.

c The experimental setup for the permeability test. Three beakers, each filled with 100 ml of deionized water, were covered by MEAP, MEAP (punctures), and polyimide, respectively. Each beaker was secured with a rubber band to ensure that water only passed through the cover. The three beakers were placed in a programmable temperature and humidity tester (QHP-360BE, LICHEN, China) set to a temperature of 33 ℃ and a humidity of 30%, simulating the conditions on human skin.

d The water loss rates in the three beakers. The MEAP (punctures) exhibited higher water loss compared to the MEAP, indicating that the permeability can be adjusted by modifying the physical structure of the substrate. Measurements were recorded for each beaker every hour, with n = 3 samples for each recording.

e The water loss rate of each cover. Considering that the insensible sweat rate of individuals ranges from 12 to 42 $g \cdot m^{-2} \cdot h^{-1}$ [2], the permeability of the MEAP is sufficient to provide a comfortable wearing experience for daily use.

3. For the reader's better understanding, more details are expected on the MEAP connected to the data acquisition module via flexible printable circuit board connectors.

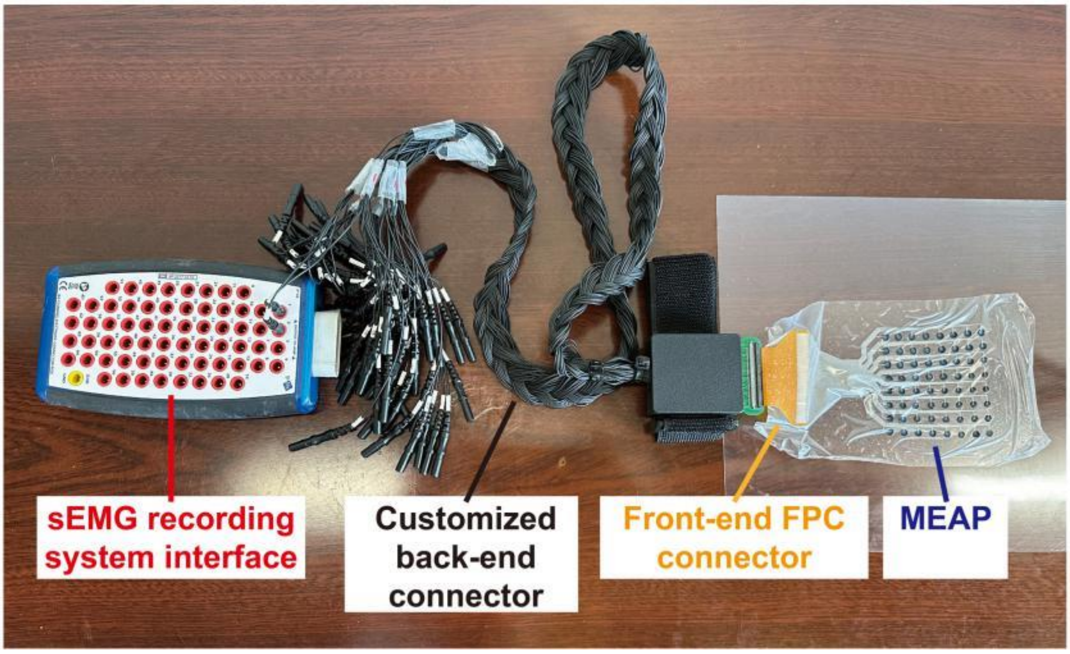
Our response: We appreciate the reviewer's request for more details regarding the connection. In our study, we employed hot-pressing to combine the Flexible Printed Circuit (FPC) with the MEAP. By using a customized back-end connector, each channel of the MEAP could be independently connected to the G.tec recording system. For better understanding, we have included a photograph in the Supplementary Information, illustrating the entire setup.

Supplementary Fig. 20 The whole setup for connection between MEAP and EMG recording system.

See page 16:

The MEAP was connected to the sEMG recording system via flexible printed circuit (Supplementary Fig. 20).

We also added more information in the MATERIALS AND METHODS. See page 32:



Note that electrode sites and connection pads were protected by silicone films during the encapsulation to allow the electrodes and connection pads to be exposed.

The front-end connectors were made by polyimide flexible printed circuit (FPC). Front-end connectors were designed and fabricated (EasyEDA, China) for connecting specific MEAPs. The FPC and MEAP were hot-pressed together with force of 50 N and temperature of 140 °C for 30 s by a hot-pressing machine (G311, Freamc, China). With a customized back-end connector, every channel of MEAP can be independently connected to EMG recording system.

*4. The authors compared the attachment performance between the commercial array and MEAP. In fact, the spatial density and distribution of the electrode sites is also important to its application. The commercial array has 64 channels with 8*8 array while the MEAP is 24 with 4*6 array, which leads to mismatch location of the electrode sites during comparison. It would be better to make a comparison directly with the same configuration.*

Our response: We appreciate the reviewer's insightful comment, and we highly value the advice provided. We agree with the reviewer's suggestion regarding the configuration of the electrode array and its potential impact on the recording results. To ensure a fairer comparison, we repeated experiment using the updated 64-channel MEAP. This updated MEAP has the exact same configuration as the commercial electrode array, including the surface area of the substrate (Supplementary Fig. 19). We applied the same analysis method to process the data obtained from the experiment, and the results consistently demonstrated that the 64-channel MEAP exhibited superior and more stable performance in terms of baseline noise and SNR when compared to the commercial electrode array, both on muscle and muscle-tendon junction recordings. Although the smaller contact area of the electrode on the 64-channel MEAP led to a slightly lower SNR compared to the previous 24-channel MEAP, the differences in performance between the MEAP and the commercial electrode array remained statistically significant. Given these findings, most of our previous conclusions remain valid, and we have made minimal changes to the text. The main changes have been implemented in Figure 5 and two Supplementary Videos.

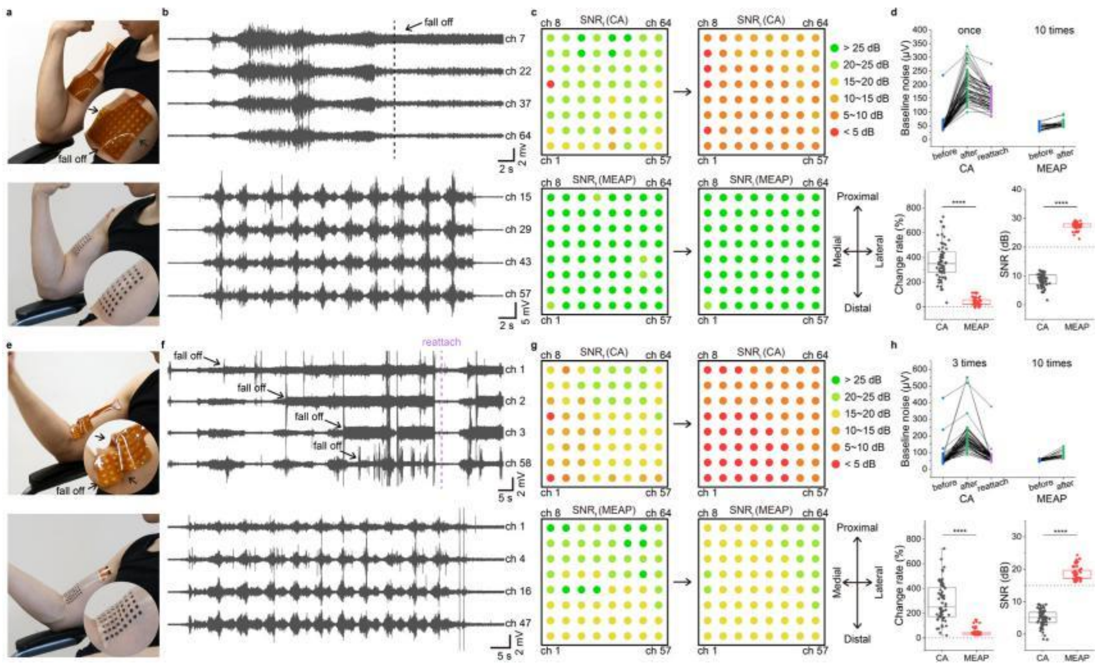


Fig. 5 The comparison of attachment performances between commercial array and MEAP.

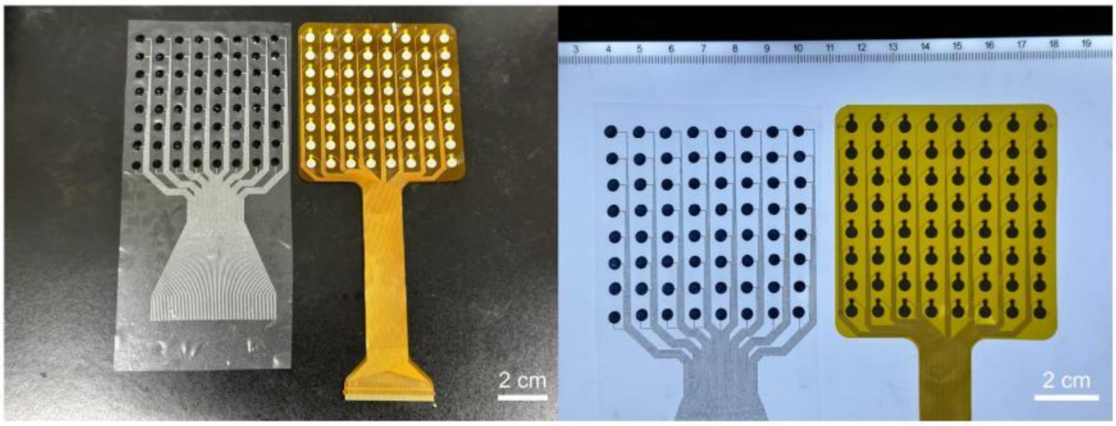
a, e Photographs of CA and MEAP attachment, illustrating the difference when recording from muscle and muscle-tendon junction of biceps brachii.

b, f sEMG signals recorded using CA and MEAP on muscle and muscle-tendon junction of biceps brachii. Four typical channels were picked for each recording.

c, g Spatial SNR performance map for each channel of CA and MEAP for the first and last muscle contraction. SNR_f: SNR of the first contraction; SNR_l: SNR of the last contraction.

d, h Statistical analysis of performances between CA and MEAP, including baseline noise level of CA before and after one or three muscle contractions, as well as after reattachment; baseline noise level of MEAP before and after ten muscle contractions; baseline noise change rates before and after muscle contractions; SNR performance of the last muscle contraction recorded by each of the CA and MEAP channels. Significance was determined by one sample t test (*$P< 0.05$; **$P< 0.01$; ***$P< 0.001$;

****P< 0.0001).



Supplementary Fig. 19 The configuration comparison between MEAP and CA. Both arrays have electrode diameter of 4 mm and IED of 8 mm.

See page 16: ...with human skin (Young's modulus of 10 kPa). To fairly compare the contact performance on the skin, we fabricated a 64-channel MEAP with the same configuration as the CA (Supplementary Fig. 19). We recorded a movie to ... resulting in a significantly lower SNR. While the MEAP exhibited a much more stable noise level even after ten contractions, maintaining a high SNR. We also recorded sEMG signals... the first contraction to the last one (Fig. 5g). MEAP on the other hand, produced stable recordings with all channels having SNR greater than 15 dB. Statistical analysis of sEMG...

5. In addition to the innovative aspects of the PPT electrode and the MEAP, please also specify the corresponding limitations at Conclusion Section.

Our response: We appreciate the reviewer's suggestion for a more comprehensive discussion of our work. We acknowledge that there are certain limitations to the current MEAP design, and we are actively working towards addressing them to expand the potential applications of MEAP. One specific area of focus is the accurate recording and recognition of single motor unit signals from surface electromyography (sEMG) signals. If MEAP can achieve this capability, it has the potential to be utilized in clinical diagnosis, replacing the use of needle electrodes. This would offer patients a much more comfortable experience while maintaining the accuracy and reliability of the diagnostic process. Such a development could revolutionize EMG clinical diagnosis by replacing invasive tools with non-invasive alternatives. Furthermore, we recognize that the cable connection is currently a limiting factor in the daily application of MEAP. We are actively addressing this issue to improve the overall user experience and make MEAP more suitable for everyday use. We have incorporated these discussions into the conclusion section of our manuscript to provide a more comprehensive overview of the potential implications and future directions of our work.

See page 30: e.g., prosthetics or virtual reality. We also aim to replace invasive needle electrodes in clinical applications with MEAP to provide patients with improved comfort. However, current MEAPs have difficulty recognizing single motor unit signals from sEMG recording because the lack of intelligent back-end algorithms causes low single motor unit selection efficiency and accuracy. In this case, MEAP can only provide limited help to clinical diagnosis. Another drawback of current MEAPs is they are using cable connection, which limits the application scenarios and simplicity. To address these issues, we are endeavoring to combine MEAP with intelligent algorithms and wireless modules to make whole devices more useful in clinic scenario and more portable in daily life. We believe in the future, MEAP has enormous potential to be commercialized because of its low cost and simple fabrication, thus providing a new platform for disease diagnosis, daily rehabilitation management and scientific exercise.

Reviewer #3 (Remarks to the Author):

The authors have tried to address some of my concerns. However, they only succeeded to a very limited extent. The manuscript credibly demonstrates that the mechanical properties of the new electrode array are superior to other electrode arrays. The same applies to the purely electrical properties.

Our response: We sincerely appreciate the reviewer's acknowledgement of our effort and the recognition of the superior mechanical and electrical properties of our electrode array. We are grateful for the time and attention the reviewer dedicated to carefully reviewing our manuscript once again.

However, the manuscript still has major shortcomings when the suitability of the new electrode array is examined with regard to the detection of sEMG signals and compared with state-of-the-art methods. As an example, I refer again to Fig. 4e, which still compares the sEMG signals derived with the new electrodes with signals from sEMG derivations that do not comply with the international recommendations for the derivation of sEMG signals. The international recommendations for the detection of sEMG signals are defined and described in the SENIAM and CEDE projects. The project results are published as well as described on the internet. E.g. it seems, that the electrode distance is larger than 1/4 of the muscle fibre length of the FCU and that the sensor is placed halfway the (most) distal motor endplate zone and the distal tendon (SENIAM Recommendations).

Our response: We thank the reviewer for the criticism about the sEMG derivation in our manuscript. We appreciate your feedback as it helps us improve the article. Upon thorough examination of our sEMG derivation method and the articles the reviewer suggested, we found that the majority of our protocols align with international recommendations. Allow us to address each point in detail:

We have carefully compared two standards proposed by the reviewer with our electrodes based on the SENIAM guidelines. We apologize for the omission of the Inter-electrode distance (IED) information in the 'sEMG signal recording' section, where the IED between two Red Dot 2223 or TPP electrodes is consistently set at 20 mm. This oversight may have led the reviewer to perceive that '*the electrode distance is larger than 1/4 of the muscle fiber length of the FCU*'. However, after reviewing several articles investigating the muscle length of the Flexor Carpi Ulnaris (FCU), we found that our IED of 20 mm is actually less than 1/4 of the typical muscle fiber length[3,4]. For instance, Table 1 in Ref 3 demonstrates that the FCU muscle length is '***236.5 ± 5.4 mm***'.[3] Additionally, Ref 4 provides the details that '***The flexor carpi ulnaris muscle presents a total length of 26.5 cm, with a muscular belly of 24.5 cm of length by 3.5 cm of width and 0.5 cm of thickness;***'[4] These findings confirm that the usual length of the FCU exceeds 200 mm, indicating that our IED of 20 mm falls within the acceptable range of being less than 1/4 of the muscle fiber length of the FCU.

Regarding electrode placements, SENIAM recommends the sensor be placed halfway between the (most) distal motor endplate zone and the distal tendon. However, the endplate zone is hard to identify for each muscle, so researchers usually attach the electrodes on the muscle belly or the bulkiest part of the muscle[5], which is also reported by SENIAM. We opted to follow this guidance because we did have difficulty in locating the endplate zone precisely. We supplement this information in our method section.
See page 34: 'For fatigue comparison tests, electrodes were attached on the most prominent bulge of the muscle belly of FCU, and different types of electrodes were attached on the exactly same position.'

To provide further clarification that we followed the SENIAM recommendations for all our sEMG derivations, we present the evidence below:

Electrode Shape and Size: As we compared our electrodes with the commercial Red Dot 2223 electrodes, the shape and size parameters were predetermined. However, we have previously discussed the reliability of these electrodes in our response letter, ensuring that this aspect aligns with the standard.

Inter-Electrode Distance (IED): '***SENIAM recommends to apply the bipolar SEMG electrodes around the recommended sensor location with an inter electrode distance***

of 20 mm.' We have followed the recommendation and adjusted the IED to 20 mm accordingly.

Orientation of Electrodes: '***SENIAM recommends that the bipolar SEMG electrodes should be placed around the recommended sensor location with the orientation parallel to the muscle fibres.***' We have carefully followed the recommendation and aligned the orientation of our electrodes parallel to the FCU fiber.

Fixation on the Skin: ***'SENIAM recommends to use elastic band or (double sided) tape / rings for the fixation of the electrodes(construction) and cables to the skin in such a way that the electrodes are properly fixed to the skin, movement is not hindered and cables are not pulling the electrodes(construction).'*** Our Ag/AgCl and TPP electrodes were properly fixed to the skin because they are both sticky.

Location of the Reference Electrode: ***'Depending on the application SENIAM recommends to use the wrist, the proc. spin. of C7 or the ankle as the standard location of the reference electrode.'*** In our study, we opted to attach the reference electrode to the elbow to maintain a stable reference potential.

With regard to CEDE, we inspected all related published matrices[6–10] but could not find any details about sEMG derivations. Instead, we used the article[11] by Hermens et al (https://doi.org/10.1016/S1050-6411(00)00027-4, with >6000 citations on Google scholar) which is cited by the CEDE matrices and provides details about sEMG derivations, to compare with our own sEMG recording protocols. Allow us to compare each point in detail:

Electrode material: '***For bipolar or monopolar electrodes, it is obvious that Ag/AgCl was the preferred electrode material.***' ***'It is recommended to use pre-gelled Ag/AgCl electrodes.'*** In our manuscript, the Red Dot 2223 electrodes (pre-gelled Ag/AgCl electrodes) were used.

Electrode shape and size: ***'Thus, in the literature both rectangular (bars) and circular electrodes are being used for SEMG recordings of which circular electrode are by far the most used.'*** In our manuscript, the Red Dot 2223 electrodes are pre-gelled and have a surface area of 2.4 cm^2. To make a fair comparison, our TPP electrodes fabricated with the same surface area were used.

Inter-electrode distance: ***'Authors seem to have a preference for IED values which are a multiple of 10 mm. The largely preferred distance was 20 mm.'*** We have followed the recommendation and adjusted the IED to 20 mm accordingly.

Skin preparation: ***'In the remaining 76 (53%) publications, standard skin preparation techniques were mentioned such as shaving, rubbing/abrasion and cleaning of the skin, or a combination of these techniques.'*** In fact, one advantage of our TPP electrodes is that they require no skin preparation or gel, as their soft adhesive properties allow excellent contact with the skin, even in the presence of hair. Although feasible, we did not use skin preparation or gel for both the commercial Ag/AgCl and TPP electrodes to ensure a fair comparison. We specifically selected a less-hairy position on the arm, which is the FCU, to obtain higher signal quality.

Sensor location and orientation on the muscle: ***'Globally, three placement strategies can be discerned: 1. on the center or on the most prominent bulge of the muscle belly (10 out of 21); 2. somewhere between the innervation zone and the distal tendon (6 out of 21); 3. on the motor point (1 out of 21).'*** Compared to SENIAM, the most prominent bulge of the muscle belly was also mentioned for electrode placement in this article. This is the exact location we selected for our comparison tests between Ag/AgCl and TPP electrodes on FCU, because we are not able to figure out the precise location mentioned in the second strategy. And the first strategy (on muscle belly) was most used by other researchers, making us believe our sEMG derivations accords with international standards.

In our later sEMG derivations using the MEAP, we adopt a high-density recording approach instead of the conventional bipolar recording method. This approach allows us to capture sEMG signals using our novel array patch, which is not currently available on the market. As a result, it may not always be applicable or appropriate to utilize established standards designed for conventional tools when designing protocols for our unique tool. But we still follow the fundamental mechanism of sEMG and some high-density recording rules[12] to set our protocols, such as the configuration of array, the attachment position on the muscle and particular tasks for sEMG recording. We believe the reviewer will understand and appreciate our effort.

In addition, any information about the sensor used, such as size of the active area, interelectrode distance and exact position of the electrodes as well as the treatment of the skin, is missing. This information is essential to evaluate the quality of the derived sEMG signals and is standard in any sEMG publication, even when comparing two methods

Our response: We thank the reviewer asking more details about all EMG recording process. Upon reevaluation of our manuscript, we have identified that some of the information, although included, was not presented clearly. We agree with the reviewer that this information is critical for sEMG publication, and we have summarized these below.

Size of the active area: we mentioned them in the experiment section **'Impedance measurement'** on page 34, but we also found the description in **'sEMG signal recording'** is missing, so we added the information to this part.
See page 34: '2228 and 2223 electrodes have the surface contact area of 2 and 2.4 cm^2. The TPP electrodes in each comparison test have the same surface contact area with 2228 or 2223 electrodes correspondingly.'

Interelectrode distance:

We added the detailed information to the '**sEMG signal recording**' part.

See page 34: 'All interelectrode distance for bipolar recording is 20 mm.'

Position of the electrodes:

We added the detailed information to the '**sEMG signal recording**' part.

See page 34: 'Foam Monitoring 2228 electrodes were used for long-term test, flexibility test on the forehead, and Red Dot 2223 (USA, 3M) electrodes were used for fatigue comparison tests. For fatigue comparison tests, electrodes were attached on the most prominent bulge of the muscle belly of FCU, and different types of electrodes were attached exactly on the same position.'

Treatment of the skin:

As mentioned in the previous part, no needed treatment of the skin is an advantage of our electrodes. Thus, we did not use any treatment of the skin in all sEMG tests through our manuscript.

See page 35: 'For all sEMG recording, no skin treatment was used, including shaving, rubbing or cleaning of the skin.'

We thank again the reviewer helping us supplement more detailed information to the manuscript for readers to understand clearer.

Additionally, open cable clips were used to fix the cables to the electrodes. This worsens the SNR considerably. Shielded connectors are more common.

Our response: We acknowledge the reviewer's viewpoint that shielded connectors are more commonly used and preferable for eliminating factors that may degrade the SNR. However, we encountered difficulties in finding a shielded connector on the market that would be compatible with our single-channel TPP electrodes. Developing a custom shielded connector would have been costly and time-consuming. As a compromise, we opted to use crocodile clips as connectors and employed tape fixation to mitigate SNR reduction as much as possible. It is worth noting that several articles published in reputable journals have also used open cables, even at the expense of sacrificing SNR values[13–15]. In this comparison test, our objective is to assess the performance of the electrodes rather than solely pursuing the highest SNR. Thus, ensuring a fair comparison between the two electrode types is our primary focus. As a result, we have used open cables for both the Ag/AgCl electrodes, based on the same connector used for the TPP electrodes. This decision was made to ensure consistency and fairness in our comparison. We hope that the reviewer understands our rationale behind using open cables for both electrode types.

The manuscript is still full of claims - often in the superlative - about signal quality and possible applications of the described electrode, which are not statistically proven. I would like to draw special attention to the statistics here once again! I do not see any statistical calculation in the entire manuscript that shows that the claims made with respect to application are statistically significant, or that at least a tendency can be read. Instead, individual examples are shown, which is good, but does not justify the strong formulations and the emphasis on the new electrode over existing methods. That the major problem with the paper is the lack of statistics.

Our response: We appreciate the reviewer for bringing up the lack of statistical analysis in our manuscript. We carefully reviewed our text and found that there was only one instance of a superlative expression found on page 9: "***In the comparison, we also found only 6 out of 13 studies discussed sticky electrodes, and our TPP electrodes perform the best in terms of adhesiveness, which is an important contribution to its highest SNR among all dry electrodes.***" We provided Figure 2**l** as supporting evidence, and all references can be found to support this statement. Thus, we believe that the superlative expression used in this context is appropriate. With regard to the signal quality comparison, we agree with the reviewer's suggestion that the inclusion of statistical analysis would strengthen our results. We have now incorporated statistical analysis into Figure 5**d** and **h**. This addition provides more robust evidence to support the conclusion that MEAP demonstrates superior performance compared to commercial arrays in terms of signal quality and stability.

See page 18:

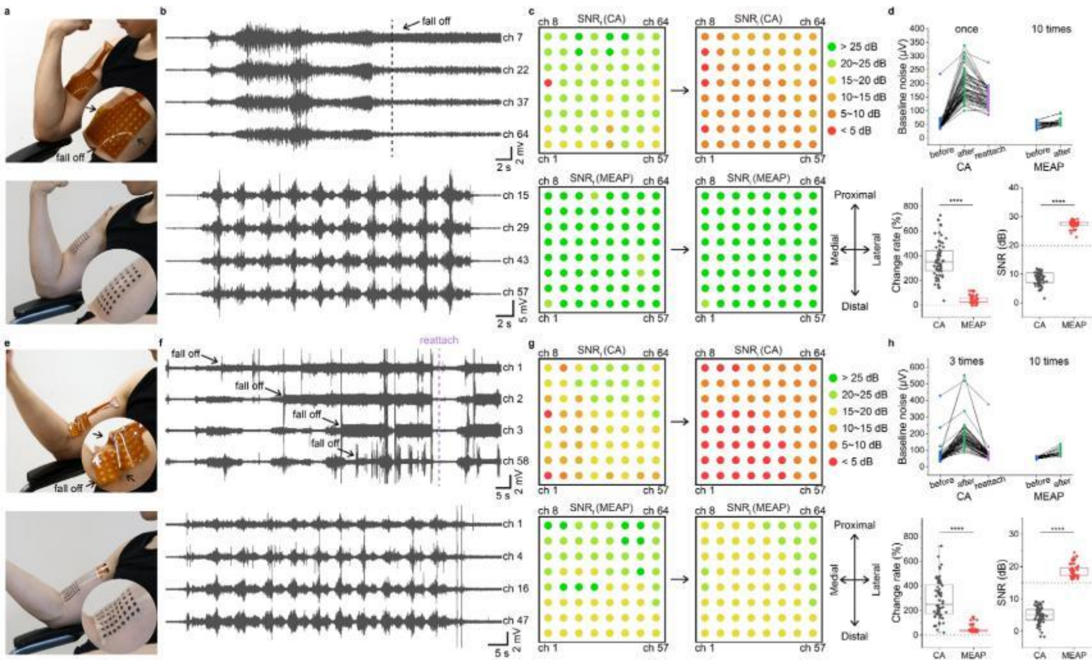


Fig. 5 The comparison of attachment performances between commercial array and MEAP.

a, e Photographs of CA and MEAP attachment, illustrating the difference when recording from muscle and muscle-tendon junction of biceps brachii.

b, f sEMG signals recorded using CA and MEAP on muscle and muscle-tendon junction of biceps brachii. Four typical channels were picked for each recording.

c, g Spatial SNR performance map for each channel of CA and MEAP for the first and last muscle contraction. SNR_f: SNR of the first contraction; SNR_l: SNR of the last contraction.

d, h Statistical analysis of performances between CA and MEAP, including baseline noise level of CA before and after one or three muscle contractions, as well as after reattachment; baseline noise level of MEAP before and after ten muscle contractions; baseline noise change rates before and after muscle contractions; SNR performance of the last muscle contraction recorded by each of the CA and MEAP channels.

Significance was determined by one sample t test (*P< 0.05; **P< 0.01; ***P< 0.001; ****P< 0.0001).

We also added statistical information into the caption of Fig. 8 and 'Materials and Methods' section.

See page 28:

... by 6 channels in column 1 of MEAP.

f RMS values of sEMG signals against time (left), in the isometric task (middle) and in the dynamic task (right) across selected channels. n = 54 RMS values per channel. The box plots show the mean (center square), median (center line), the 25th to 75th percentiles (box) and the smallest and largest value that is ≤ 1.5 times the interquartile range (the limits of the lower and upper whiskers, respectively)

see page 37:

Statistical analysis:

Data are presented with mean values ± SD, unless otherwise noted in the figure caption. Significance was defined as *P < 0.05; **P < 0.01; ***P < 0.001; ****P < 0.0001. Statistical analysis was performed using Origin Pro 2021.

We also have other statistical analysis in supplementary information, including Supplementary Fig. 26-28. Moreover, to explore the potential applications of our MEAP, we performed the same experiment on three subjects, and comparable results were obtained. This evidence indicates that our MEAP can be applied consistently and reliably across different individuals. We agree with the reviewer's suggestion that including more statistical analysis in this section would better demonstrate the advantages of our tools. However, as this manuscript primarily focuses on introducing a new tool rather than presenting a specific clinical application, we believe that too many samples or tests are not necessary to establish the novelty and validity of our tools. It is important to consider that conducting statistical experiments would require significant additional expenses and time. We have also observed that the level of statistical analysis in our manuscript exceeds what is typically found in other similar publications discussing novel electrodes[13–15]. Overall, while we recognize the value of additional statistical analysis, we believe that the current evidence and results are sufficient to establish the uniqueness and potential of our MEAP tool for further exploration and development.

As it stands, there is no evidence for the results in the different applications, and for me that is a no-go for a publication.

Our response:

We appreciate the reviewer's perspective on the applications of our work. However, we would like to emphasize and clarify the comprehensiveness of our validations again. In Figure 5, we have already provided evidence that the conformability of our TPP electrode is superior to that of commercial Ag/AgCl electrodes on the forehead, and the conformability of MEAP is significantly better than CA on the biceps brachii. Also, we have shown that different attachment methods significantly impact the Signal-to-Noise Ratio (SNR). Therefore, logically, repeating the same comparisons in our subsequent applications is unnecessary, as the issue of conformability arises in most areas of the body. It is worth noting that in subjects with lower body fat percentage, this issue can be more pronounced due to greater skin deformation. However, with our MEAP, we have observed excellent performance not only in signal quality but also in RMS recording, fatigue recording, and tendon displacement. As a result, this tool is expected to exhibit superior performance in most sEMG recordings where skin deformation occurs (which is prevalent across the body). Furthermore, the successful observations of RMS, fatigue, and tendon displacement enable the application of this tool in muscle injury prevention. Because monitoring these parameters can help control tendon length, which is crucial in preventing tendon tears—a common cause of injury. It is important to clarify that our intention in this article is not to establish clinical criteria, but rather to show the capabilities of this tool and its potential applications based on those capabilities. However, we would be delighted to conduct further clinical studies with statistical data using our tools in our future studies. Considering the aforementioned characteristics, we firmly believe that MEAP brings innovation to the current tool market, aligning with the standards of the journal.

Furthermore, some of the claims about the state of the art of existing sEMG electrode/lead methods are simply wrong. Let me cite the abstract: "However, current sEMG electrodes do not offer adequate high-quality data for their widespread use in clinics and everyday life, since these are neither stretchable nor arrayed". This sentence is simply wrong! Electrode arrays have been available for many years and some of them are stretchable. Whether stretchability improves the clinical significance of the sEMG is unknown, while the clinical benefit of the array is proven. The manuscript is full of such examples and should be formulated in a less absolute way to reflect reality.

Our response:

We appreciate the reviewer comments on our inappropriate phrasing. We changed our abstract as the reviewer advice.

See page 1:

'Surface electromyography (sEMG) can provide multiplexed information about muscle performance. If current sEMG electrodes are stretchable, arrayed, and able to be used multiple times, they would offer adequate high-quality data for continuous monitoring. The lack of these properties delays the widespread use of sEMG in clinics and in everyday life. Here, we address these constraints by design of an adhesive dry electrode using tannic acid, polyvinyl alcohol, and PEDOT:PSS (TPP). The TPP electrode offers superior stretchability (~200%) and adhesiveness (0.58 N/cm) compared to current electrodes, ensuring stable and long-term contact with the skin for recording (>20 dB; >5 days). Additionally, we developed a metal-polymer electrode array patch (MEAP) comprising liquid metal (LM) circuits and TPP electrodes. The MEAP demonstrated better conformability than commercial arrays, resulting in higher signal-to-noise ratio and more stable recordings during muscle movements. Manufactured using scalable screen-printing, these MEAPs feature a completely stretchable material and array architecture, enabling real-time monitoring of muscle stress, fatigue, and tendon displacement. Their potential to reduce muscle and tendon injuries and enhance performance in daily exercise and professional sports holds great promise.'

To address the concern raised by the reviewer, we conducted a thorough examination of our manuscript and made appropriate changes. We specifically focused on ensuring that all the conclusions presented in the "Results" section were supported by our experimental findings, thereby minimizing any subjective aspects. Consequently, we have also revised the introduction and discussion sections to further enhance the clarity and objectivity of our work.

See page 2: 'However, there is very little research using sEMG techniques to make such tendon identifications.'

See page 29: 'However, it is extremely hard for commercial hydrogel electrodes to accomplish the same recording sites in the same area as MEAP.'

With regard to reviewer's comments about the clinical significance of stretchability, we agree that its direct clinical impact has never been examined. However, it is precisely because of this uncertainty, that we are utilizing our new tool, which has demonstrated clear advantages in terms of conformability and signal quality during movement, to explore its potential applications in clinical scenarios. Throughout the manuscript, we have utilized phrases such as 'we believe,' 'it would be,' 'had great potential,' and 'in the future' to emphasize that we are not claiming immediate superiority over existing clinical tools but rather aiming to facilitate further explorations based on our findings.

In summary, we greatly appreciate the reviewer's insightful comments as they have significantly improved the manuscript.

Reference

1. Cheng, J. *et al.* Wet-Adhesive Elastomer for Liquid Metal-Based Conformal Epidermal Electronics. *Adv Funct Mater* 2200444 (2022) doi:10.1002/adfm.202200444.
2. Zhong, B., Jiang, K., Wang, L. & Shen, G. Wearable Sweat Loss Measuring Devices: From the Role of Sweat Loss to Advanced Mechanisms and Designs. *Advanced Science* **9**, (2022).
3. Fridén, J., Lovering, R. M. & Lieber, R. L. Fiber length variability within the flexor carpi ulnaris and flexor carpi radialis muscles: Implications for surgical tendon transfer. *Journal of Hand Surgery* **29**, 909–914 (2004).
4. Campos, D., Bartholdy, L. M. & Souza. *Case report Accessory flexor carpi ulnaris muscle: a case report of a rare variation in human*. *J. Morphol. Sci* vol. 27 (2010).
5. Mesin, L., Merletti, R. & Rainoldi, A. Surface EMG: The issue of electrode location. *Journal of Electromyography and Kinesiology* **19**, 719–726 (2009).
6. Martinez-Valdes, E. *et al.* Consensus for experimental design in electromyography (CEDE) project: Single motor unit matrix. *Journal of Electromyography and Kinesiology* **68**, 102726 (2023).
7. Gallina, A. *et al.* Consensus for experimental design in electromyography (CEDE) project: High-density surface electromyography matrix. *Journal of Electromyography and Kinesiology* **64**, 102656 (2022).
8. Besomi, M. *et al.* Consensus for experimental design in electromyography (CEDE) project: Amplitude normalization matrix. *Journal of Electromyography and Kinesiology* **53**, 102438 (2020).
9. Mcmanus, L. *et al.* Consensus for experimental design in electromyography (CEDE) project: Terminology matrix. *Journal of Electromyography and Kinesiology* **59**, 102565 (2021).
10. Besomi, M. *et al.* Consensus for experimental design in electromyography (CEDE) project: Electrode selection matrix. *Journal of Electromyography and Kinesiology* **48**, 128–144 (2019).
11. Hermens, H. J., Freriks, B., Disselhorst-Klug, C. & Rau, G. *Development of recommendations for SEMG sensors and sensor placement procedures*. *Journal of Electromyography and Kinesiology* vol. 10 www.elsevier.com/locate/jelekin (2000).
12. Merletti, R. & Muceli, S. Tutorial. Surface EMG detection in space and time: Best practices. *Journal of Electromyography and Kinesiology* **49**, 102363 (2019).
13. Zhang, L. *et al.* Fully organic compliant dry electrodes self-adhesive to skin for long-term motion-robust epidermal biopotential monitoring. *Nat Commun* **11**, 4683 (2020).
14. Tan, P. *et al.* Solution-processable, soft, self-adhesive, and conductive polymer composites for soft electronics. *Nat Commun* **13**, 358 (2022).
15. Driscoll, N. *et al.* MXene-infused bioelectronic interfaces for multiscale electrophysiology and stimulation. *Sci Transl Med* **13**, eabf8629 (2021).

REVIEWER COMMENTS

Reviewer #2 (Remarks to the Author):

The authors have addressed my concerns in the revised manuscript. I would like to recommend its publication in Nature communications.

Reviewer #3 (Remarks to the Author):

I am tired of the discussion with the authors. Either they don't seem to understand my criticisms or, what I think is more likely, they can't address them. Two examples:

The authors use 3M RedDot 2223 and 2228 electrodes. According to the manufacturer, the RedDot 2223 has a diameter of 43.1 mm. When two electrodes are taped side by side for a bipolar lead, the smallest possible interelectrode distance (from centre to centre) is 43 mm. How is an interelectrode distance of 20 mm to be achieved with this? Even if the adhesive surface is reduced by cutting, it is difficult to maintain an interelectrode distance of 20 mm because the active electrode surface has a diameter of 16 mm. With the remaining 4 mm, sufficient adhesion cannot be achieved, which worsens the SNR. Long story short: Neither the 3M RedDot 2223 nor the 2228 meet the SENIAM standard.

Secondly: Statistic is now calculated via repetitions in individual subjects. This makes no sense at all, since the performance of the new electrode depends on where and how well it sticks. And on the other hand, it is known that the performance of different electrodes depends on the individual subject. This is due to the different skin resistance of different test persons. Even if it was tested on three subjects, that is far too few to reach a sustainable conclusion. For me, such arbitrariness does not belong in a scientific paper. The statement alone that a proper study is too time-consuming and expensive (which, by the way, is not an argument) already shows that the authors concede that a scientifically correct investigation of the question could well lead to different results.

Response to reviewers for the manuscript (NCOMMS-22-46103C)

Reviewer #2 (Remarks to the Author):

The authors have addressed my concerns in the revised manuscript. I would like to recommend its publication in Nature communications.

Our response: We truly appreciate the reviewer taking the time to carefully read our revised manuscript and provide such excellent feedback.

Reviewer #3 (Remarks to the Author):

I am tired of the discussion with the authors. Either they don't seem to understand my criticisms or, what I think is more likely, they can't address them. Two examples:

The authors use 3M RedDot 2223 and 2228 electrodes. According to the manufacturer, the RedDot 2223 has a diameter of 43.1 mm. When two electrodes are taped side by side for a bipolar lead, the smallest possible interelectrode distance (from centre to centre) is 43 mm. How is an interelectrode distance of 20 mm to be achieved with this? Even if the adhesive surface is reduced by cutting, it is difficult to maintain an interelectrode distance of 20 mm because the active electrode surface has a diameter of 16 mm. With the remaining 4 mm, sufficient adhesion cannot be achieved, which worsens the SNR. Long story short: Neither the 3M RedDot 2223 nor the 2228 meet the SENIAM standard.

Our response: We value the reviewer's time spent reading our revised manuscript and raise the concern about electrodes usage again. In Fig. 4e, we presented a photograph illustrating our method to achieve an IED of 20 mm by cutting the adhesive surface. It is important to note that both the active electrode surface and the surrounding substrate exhibit sufficient adhesiveness. Thus, the adhesive properties of the electrodes in this configuration should be adequate to ensure high-quality recording. Notably, no motion artifacts were observed in the recording depicted in Fig. 4g. Building upon our statements in last response letter regarding electrode selection, we maintain that the 3M RedDot 2223 and 2228 electrodes align with the SENIAM or CEDE standards.

Secondly: Statistic is now calculated via repetitions in individual subjects. This makes no sense at all, since the performance of the new electrode depends on where and how *well it sticks. And on the other hand, it is known that the performance of different electrodes depends on the individual subject. This is due to the different skin resistance of different test persons. Even if it was tested on three subjects, that is far too few to reach a sustainable conclusion. For me, such arbitrariness does not belong in a scientific paper. The statement alone that a proper study is too time-consuming and expensive (which, by the way, is not an argument) already shows that the authors concede that a scientifically correct investigation of the question could well lead to different results.*

Our response: To address the reviewer's concern, we modified Fig. 7, 8 and added statistics.

See page 21: This observation was also verified by palpation on the biceps brachii. Recording from five different subjects with a total of 15 MEAPs was carried out, and the mean frequency for each channel is plotted unravelling distinct trends reflecting flexion and extension, thereby demonstrating the consistent recording capability of the MEAP in identifying junction positions.

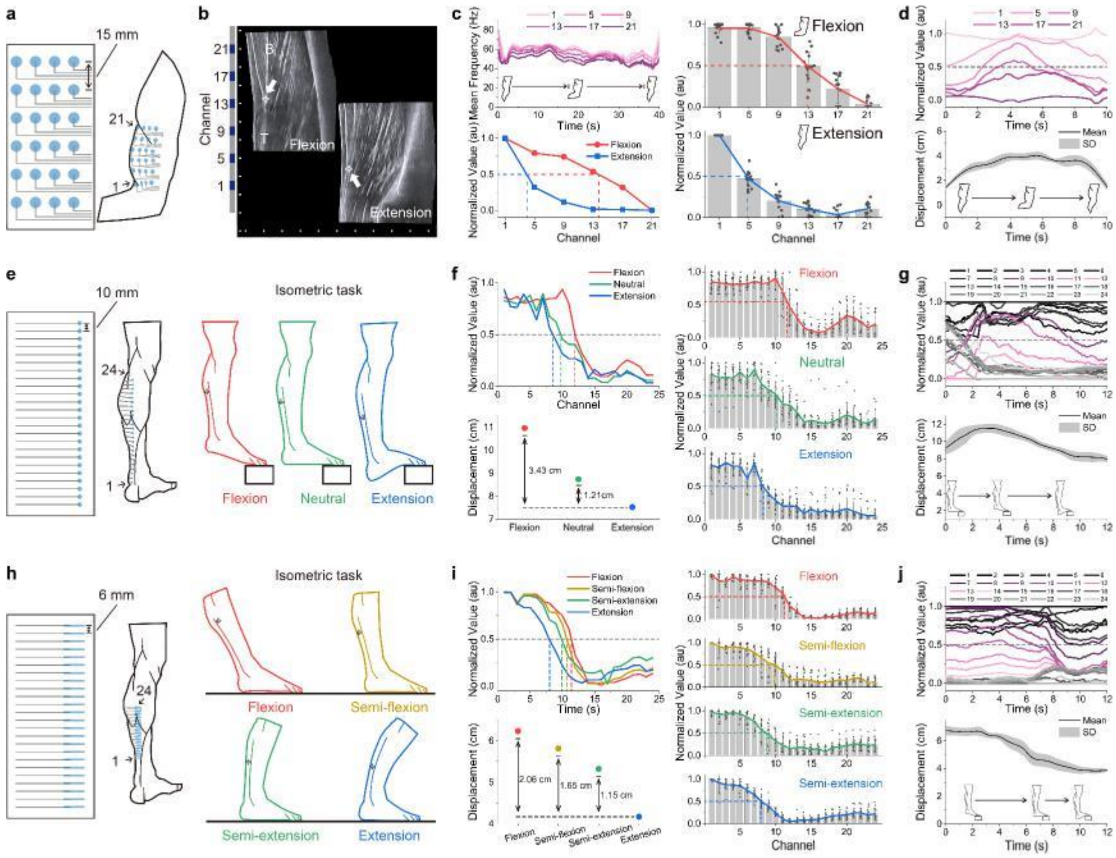

a
15 mm
21
1
b
Channel
Flexion
Extension
c
Mean Frequency (Hz)
Time (s)
Normalized Value (au)
Channel
Flexion
Extension
d
Normalized Value (au)
Time (s)
Displacement (cm)
Mean
SD
e
10 mm
24
1
Isometric task
Flexion
Neutral
Extension
f
Normalized Value (au)
Channel
Displacement (cm)
3.43 cm
1.21cm
Flexion
Neutral
Extension
g
Normalized Value (au)
Time (s)
Displacement (cm)
Mean
SD
h
6 mm
24
1
Isometric task
Flexion
Semi-flexion
Semi-extension
Extension
i
Normalized Value (au)
Time (s)
Displacement (cm)
2.06 cm
1.65 cm
1.15 cm
Flexion
Semi-flexion
Semi-extension
Extension
Channel
j
Normalized Value (au)
Time (s)
Displacement (cm)
Mean
SD

Fig. 7 Location of muscle-tendon junction by MEAP.

a Schematic diagram of a MEAP on the biceps. The IED was 15 mm. Channel numbers (1 – 24) were ordered from left to right and from bottom to top.

b Ultrasound image of tendon displacement during the isometric task with load of 5 kg and MEAP relative position on the skin. Scale bar: 1 cm.

c Mean frequencies of the EMG signals; left panels show data from the biceps brachii of a representative subject during the isometric task and the normalised mean frequencies for each channel during flexion and extension. Right panels show the normalised mean frequency data in multiple subjects. MEAPs were attached on comparable positions on the biceps brachii muscles of the subjects to obtain junction locations.

d Normalised mean frequencies of the EMG signals recorded from the biceps brachii of a representative subject during the dynamic task and real-time junction displacement in multiple subjects.

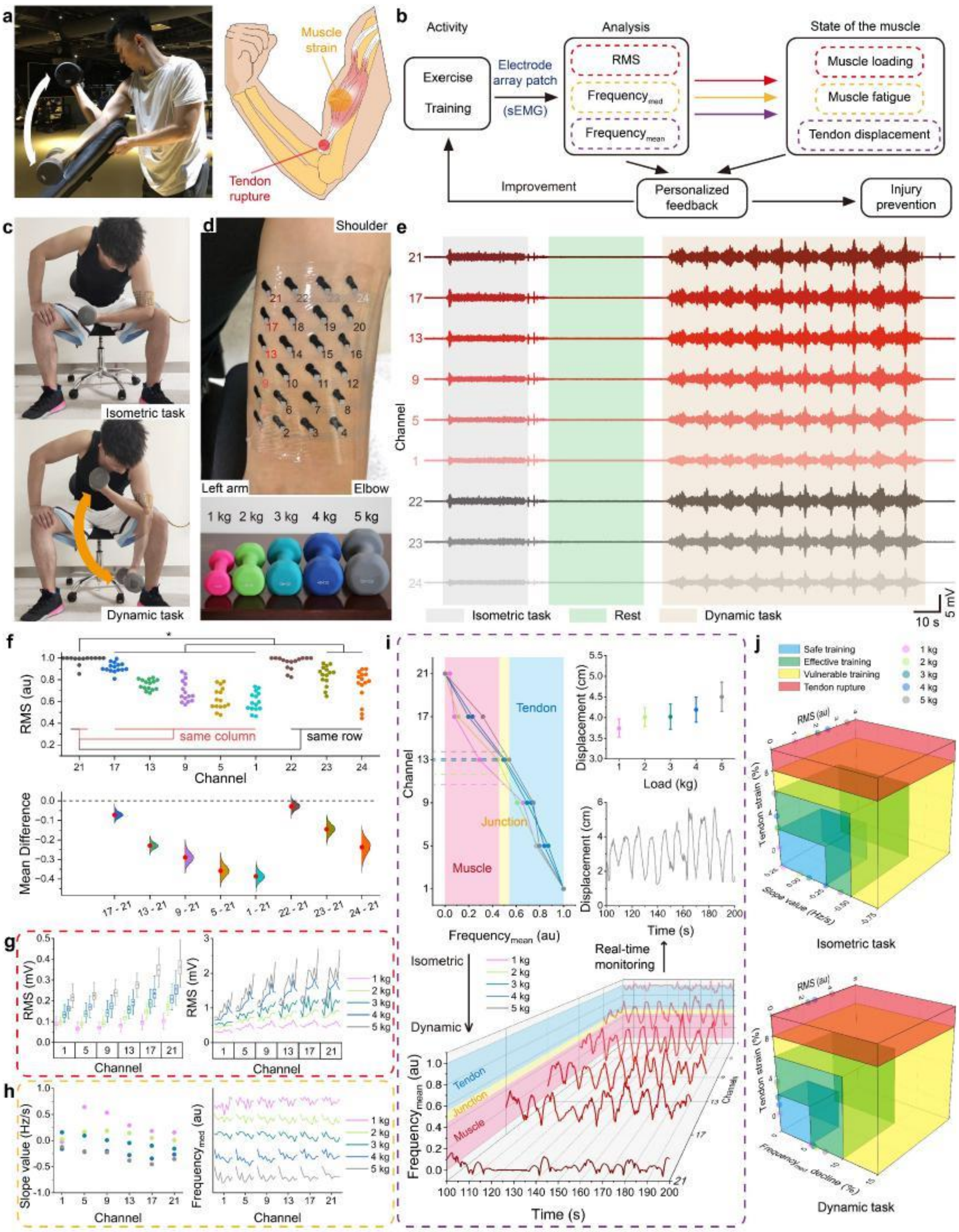
e, h Schematic diagrams of MEAPs on the gastrocnemius and Achilles tendon, and isometric tasks on a step and on the ground, respectively; the IEDs were 10 mm and 6 mm respectively. Channel numbers (1 – 24) were ordered from bottom to top

f, i Normalised mean frequencies of the EMG signals; left panels show data from the gastrocnemius and Achilles tendon of a representative subject during different isometric tasks and their corresponding displacements. Right panel shows the normalised mean frequency data in multiple subjects. MEAPs were attached on similar locations on the Achilles tendon of subjects to obtain junction locations.

g, j Normalised mean frequencies of the EMG signals recorded from the gastrocnemius and Achilles tendon of a representative subject during the dynamic task and real-time junction displacement in multiple subjects.

All displacements represent the distance between the first channel position and the junction position; all statistical experiments were conducted with the 3 repeated isometric or dynamic tasks performed by 5 subjects (n=15, different MEAPs used); mean value is represented as bars and SD means standard deviation.

Also see page 25: To verify if MEAP can provide such multiplexed information, sEMG was recorded from the biceps on 5 subjects during 5 sessions. Taking subject A as an example, each session included isometric and dynamic tasks, with load from 1 to 5 kilograms (Fig. 8c, d and supplementary Fig. 23). The sEMG recorded from the 6 channels in the left column and 4 channels in the top row were used for further detailed analysis due to their highest RMS values compared to other columns or rows (Fig. 8e). The data was statistically verified to confirm that each channel on the MEAP recorded distinct sEMG information from the muscle (Fig. 8f). Additionally, it was observed that the activation patterns of the biceps muscle are consistent even among 5 different subjects. It is worth noting that the variability of data increases as the distance between the recording channel and the control channel grows. We speculate that this phenomenon is primarily attributed to variations in muscle length among subjects. Subsequently, for a more comprehensive examination of the sEMG signals captured by MEAP, we proceeded with data analysis focused on the recordings obtained from subject A. In the isometric task, RMS amplitude increased with increasing load (Fig. 8g). ...



... by 6 channels in column 1 of MEAP.

f Statistical analysis of sEMG signals recorded by MEAP on different subjects. The Gardner-Altman plot illustrates the RMS values of sEMG signals captured by MEAPs (n=15, 3 repeated isometric tasks performed by 5 subjects, different MEAPs used). The RMS values are normalized to their respective maximum values, and channel 21 (control) is compared with others. Significance was determined by one sample t test (*P< 0.05).

g RMS values of sEMG signals...

j A visual representation of the potential for muscle injury index, generated based on the assessments made using the MEAP for isometric and dynamic task. The assessment is presented as a unified model using the measures obtained from **g-i**. The loads were classed as safe (< 3 kg), effective (3-5 kg) and vulnerable (>5 kg) based on the subject's previous experience.

Some other changes:

We removed **Supplementary Fig. 31 Summary of muscle information of three subjects** since its content has been demonstrated in Fig. 8.

To give more details, we added Data analysis section in Materials and Methods.

Data analysis:

All RMS, median frequency, and mean frequency values of the recorded sEMG signals were computed for time steps of 0.125 seconds, unless specified otherwise. For dynamic tasks in muscle-tendon junction location section, the mean frequency values of sEMG data were initially smoothed using a Savitzky–Golay filter (with a frame length of 21 and an order of 1); for real-time monitoring of dynamic tasks in the same section, each set of values were determined first and then the means were generated and plotted as mean ± SD. In the section of injury prevention, the Gardner-Altman plot was generated with a confidence level of 0.95 and a total of 5000 bootstrap samples.

We made adjustments to some sentence structures within our manuscript to enhance clarity and conciseness, while ensuring that no conclusions have been altered.

For some examples,

see page 13: Due to the excellent flexibility and adhesiveness of electrode and substrate, TPP electrodes can always make perfect attachment to the skin no matter if the skin is compressed or stretched (Fig. 4a).

See page 14: To reduce errors caused by fatigue, we linear fitted the first 25 s of each contraction to quantify the change Linear fittings were made for first 25 s of each contraction, to quantify the outcome with less errors[73].

See page 21: We verified our MEAP-based findings with ultrasound images of biceps distal tendon in a representative subject while the subject performed the isometric task with load of 5 kg (Fig. 7a, Supplementary Video 5). The positional difference of muscle-tendon junction was about 3.81 cm between flexion and extension confirmed using ultrasound image (Fig. 7b).

See page 22: We also obtained the real-time displacement of Achilles tendon junction accurately when switching from plantarflexion to dorsiflexion even in different subjects (Fig. 7g). Once the junction movement range was identified, a MEAP with shorter IED of 6 mm was used to further improve the precision of the location (Fig. 7h).

See page 27: As a result, such a high-possible injured circumstance is depicted as red range in the muscle injury index. For other three ranges, each one should be determined specifically by the exerciser under professional instructions. The MEAP successfully provided information about muscle loading, muscle fatigue and tendon displacement of the other subjects, which verified the stable and reliable recording using MEAP across all individuals (Supplementary Fig. 29, 30).