Novel Methods for Sensing Acoustical Emissions From the Knee for Wearable Joint Health Assessment

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Abstract*—*Objective: **We present the framework for wearable joint rehabilitation assessment following musculoskeletal injury. We propose a multimodal sensing (i.e., contact based and airborne measurement of joint acoustic emission) system for at-home monitoring.** ***Methods:*** **We used three types of microphones—electret, MEMS, and piezoelectric film microphones—to obtain joint sounds in healthy collegiate athletes during unloaded flexion/extension, and we evaluated the robustness of each microphone's measurements via: 1) signal quality and 2) within-day consistency.** ***Results:*** **First, air microphones acquired higher quality signals than contact microphones (signal-to-noise-and-interference ratio of 11.7 and 12.4 dB for electret and MEMS, respectively, versus 8.4 dB for piezoelectric). Furthermore, air microphones measured similar acoustic signatures on the skin and 5 cm off the skin ($\sim$4.5$\times$ smaller amplitude). Second, the main acoustic event during repetitive motions occurred at consistent joint angles (intraclass correlation coefficient $\mathrm{ICC}(1,1) = 0.94$ and $\mathrm{ICC}(1,k) = 0.99$). Additionally, we found that this angular location was similar between right and left legs, with asymmetry observed in only a few individuals.** ***Conclusion:*** **We recommend using air microphones for wearable joint sound sensing; for practical implementation of contact microphones in a wearable device, interface noise must be reduced. Importantly, we show that airborne signals can be measured consistently and that healthy left and right knees often produce a similar pattern in acoustic emissions.** ***Significance:*** **These proposed methods have the potential for enabling knee joint acoustics measurement outside the clinic/lab and permitting long-term monitoring of knee health for patients rehabilitating an acute knee joint injury.**

***Index Terms*—Biomechanics, joint sounds, wearable devices.**

I. Introduction

THE knee is one of the most complex joints in the body [1] and is thereby subject to extreme stress due to the multidirectional forces exerted on the joint during motion [1], [2]; additionally, its intricate structural arrangement, reliance on soft tissue networks for structural stability, and large loading requirements leave the joint particularly susceptible to injury [3]–[5]. As a result, the knee represents not only one of the most frequently injured body parts but also accounts for many severe injuries in terms of time of restricted and/or total loss of participation among athletes [6]–[8], military personnel [9], and other populations engaged in high performance activities [10]. Moreover, knee injuries are not exclusive to active populations; sedentary populations may be at higher risk for such injuries due to poor cardiovascular health, atrophied surrounding muscles which fail to properly stabilize the joint, and lack of training and warm-up [2], [10]. This frequency across populations combined with the extensive treatment requirements—often entailing surgery and/or substantial rehabilitation [3]—result in approximately 10.4 million patient visits annually in the United States [11]. To this extent, these injuries are considerable in their effect on not only the health care system but also on patients' daily lives given the knee's significance in performing ambulatory motions and other everyday activities [4].

To alleviate such strains on the health care system and facilitate monitoring of patients during daily activities, researchers have explored the use of wearable devices to unobtrusively acquire health information [12]. With regard to musculoskeletal and biomechanical-related disorders and injuries, these systems may provide a new way to collect objective and quantitative data. For example, Rampp *et al.* assessed gait impairment parameters in elderly populations using data from inertial sensors worn on shoes, thus providing a successful wearable alternative to clinic-based diagnostics (e.g., specialist observation, camera-based laboratories, sensor-embedded walkways, etc.) [13]. Atallah *et al.* also investigated gait, leveraging changes in gait patterns in postoperative, knee-replacement patients to evaluate recovery progress by utilizing an ear-worn sensor [14]. Toffola *et al.* developed a wearable sleeve to record robust knee joint kinematics and subject compliance during long-term, at-home activities, and therapies [15].

Moreover, acoustics can provide an unobtrusive method—and thus a possible wearable platform—for capturing information regarding *underlying* physical structures and alignments, articulating surfaces, and soft tissue characteristics. Friction between the structures and articulating components of the knee joint gives rise to various kinds of vibrations [16]. These vibrations (i.e., acoustical energy) travel to the skin surface where they encounter a large impedance mismatch between the fluid-filled tissue and air. Because of this, most of the acoustical energy

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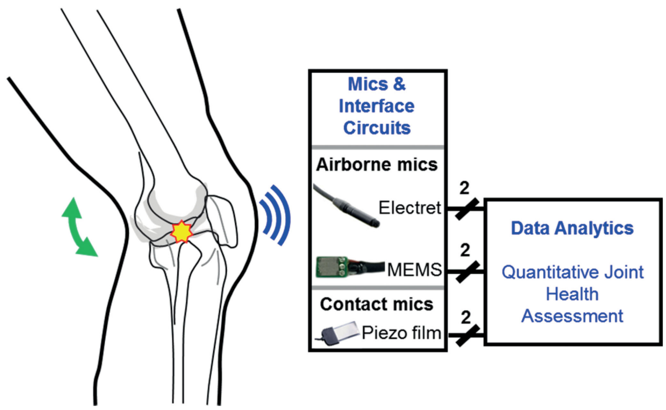

Fig. 1. Block diagram of knee joint acoustic emissions sensing and interpretation for quantifying joint health during rehabilitation.

manifests itself as vibrations signals on the skin with the majority of the energy reflected back into the tissue [17]. However, there is a small amount of energy that propagates to the air, resulting in audible joint sounds. While some very early work in this area studied airborne signals using "air" microphones, the majority of research has largely utilized vibration sensors as "contact" microphones (e.g., accelerometers, piezoelectric devices, stethoscopes) to measure joint sound vibrations [18].

Researchers have concentrated on the efficacy of joint acoustic emissions, or vibroarthographic signals (VAG), as clinically-relevant biomarkers for joint health, and notably, the majority of the research has worked toward developing diagnostic techniques to differentiate "healthy" versus "unhealthy" knee joints, primarily as it concerns cartilage-based conditions such as osteoarthritis and chondromalacia. For example, Mollan *et al.* measured acoustic emissions from the knee using a condenser microphone and captured low frequency signals ($<$100 Hz) [17] while Shark *et al.* used wide-band piezoelectric sensors to record emissions in the ultrasonic band ($>$20 kHz) [19]. They observed differences between healthy knees and those afflicted with osteoarthritis and found that osteoarthritic knees produce more frequent, higher peak, and longer duration acoustic emissions compared to healthy knees [19]. Lee *et al.* also evaluated osteoarthritic subjects using an accelerometer and successfully classified three different conditions of the patellofemoral joint [20]. To achieve such outcomes, significant work has been devoted in developing various signal processing techniques for conditioning and classifying VAG signals. Algorithms have leveraged linear prediction [21] and autoregressive modeling [20], statistical parameter investigation [22], Fourier [23] and time-frequency [24] analysis, wavelet decomposition [25], [26], and neural networks and other classifier methods such as dynamic weighted classifier fusion [22], [26].

Our ultimate goal is to enable around-the-clock monitoring of joint acoustics during normal activities of daily living, and prescribed rehabilitation activities that elicit specific signatures indicative of improving or worsening joint health. Toward this goal, we are investigating miniature sensors that can be readily integrated into a wearable device enabling, for the first time, wearable joint acoustics sensing (see Fig. 1). Our preliminary study examined possible sensors, and findings from proof-of-concept experiments suggested that the main acoustic event during repetitive motion occurs at the same angular location [27]. However, these conclusions were reached by visual observation of the signals and failed to rigidly characterize these consistencies. Thus, one main goal of this study was to quantify the consistency of main knee joint emissions with respect to joint angle position. In particular, we focused on the analysis of airborne joint sounds, which have not been extensively studied previously.

II. System Design and Methods

A. Microphone Selection

When selecting the types of microphones to include in our system, we considered 1) their ability to sense acoustic emissions and 2) their practicality for integration within a wearable system. Analysis of how joint sounds propagate through the tissue and transmit to the air suggest contact microphones are the most appropriate sensor for acquiring joint sounds, and a review of prior art [18] shows that most researchers employ contact microphones successfully in *clinical/lab* applications. Contact microphones should theoretically acquire the highest quality acoustic signal since it senses the original non-attenuated signal and is not sensitive to background noise. However, during motion and unsupervised *at-home* activity, loss of the sensor-to-skin interface is likely and of significant concern, for any compromise to the interface will be detrimental to the signal. In the extreme case that the sensor loses contact with the skin, the system will be unable to record joint sounds completely. To improve robustness, air microphones provide complementary sensing capabilities. The signal obtained by the air microphones will be inherently different from the contact microphones; the air microphones will only detect the airborne sounds: attenuated, higher frequency signals. Additionally, while not limited by the sensor-to-skin interface like contact microphones, air microphones are much more susceptible to background noise. For these reasons, we employed both sensing modalities—contact and air microphones—to more robustly capture the acoustic emissions from the joint in future implementation in a wearable device.

For the contact microphone, we selected a piezoelectric film (SDT, Measurement Specialties, Hampton, VA, USA) because its form factor seemingly lends itself to a wrap and other devices conventionally worn on the knee. Furthermore, piezoelectric films have wider bandwidths compared to miniature, low cost accelerometers, allowing for sensing of high-frequency audio signals.

Two types of air microphones were chosen to supplement the piezoelectric film in acquiring acoustic emissions from the knee joint. The first was a commercially available electret microphone (Sanken Microphone Co., Ltd., Japan). The second was a microelectromechanical systems (MEMS) microphone, specifically the MP33AB01 (STMicroelectronics, Geneva, Switzerland), which was mounted on a custom printed circuit board. Electret and MEMS microphones sense sounds in a similar manner; however, the commercial electret microphone is much more expensive ($\sim\times100$) compared to the MEMS microphone. The MEMS's low-cost and sensing capabilities provide a realistic solution for implementation in a wearable device; however, both the electret and MEMS microphones were used during our experiments with the electret microphone acting as the industry standard in terms of the quality of the sound acquired. For this study, we focused on the recordings from the air microphones because, at this time, they provide higher quality recordings as discussed in the results (Section III-A, Microphone Comparison).

B. Methods for Microphone Comparison

The similarity of the MEMS and electret microphones in detecting knee joint acoustic emissions was quantified by computing the information radius between the normalized histograms of these signals, which were acquired by both sensors at the same time placed in the same location on the lateral side of the patella. To construct the aforementioned histograms, the signals acquired from the microphones were first normalized such that their amplitudes were limited to the range $[0, 1]$. The histogram was formed from this normalized signal using 1000 bins.

Next, the quality of each sensor was evaluated by computing the signal-to-noise-and-interference ratio (SNIR). The SNIR for each microphone was calculated by finding the ratio of the peak power of a "click" (i.e., acoustic emission) emitted by the knee joint to the peak power of interface noise in the vicinity of the click. For this calculation, acoustic emissions from the microphones positioned at the medial side of the patella for the air microphones and distal side of the patella for the contact microphone were used.

Finally, a proof-of-concept experiment was conducted to compare signals measured on and off of the skin. A subject performed three cycles of seated, unloaded knee flexion/extension with two electret microphones positioned at the lateral side of the patella, one on the skin and one located 5 cm off the skin. The resulting signals were then compared.

C. Interfacing Circuits

The analog front-end for the MEMS microphones consisted of a noninverting amplifier stage with 33 dB gain, which was selected such that the signals do not saturate but are amplified to utilize the full dynamic range of the subsequent analog-to-digital converter, and a high-pass 15 Hz cutoff frequency. This stage was followed by a second-order low-pass filter with a cutoff frequency of 21 kHz. A bandwidth of 15–21 kHz was chosen, as knee joint sounds can range between these frequencies [18].

The analog front-end for the piezoelectric film microphones consisted of an amplification stage of gain 45 dB and 100 Hz high-pass cutoff. This stage was followed by a fourth-order low-pass filter with a 10 kHz cutoff frequency. A 100 Hz high-pass cutoff was chosen to attenuate the interface and motion artifact noise.

D. Human Subject Study and Measurement Protocol

Thirteen male subjects without history of knee injuries participated in the study and gave written informed consent approved by the Georgia Institute of Technology Institutional Review Board and the Army Human Research Protection Office. The subject population was reasonably homogenous in terms of physical activity level (collegiate athletes) and ranged in age (19–21 years), weight (84.1–135.3 kg), and height (174–195 cm). With this approach, our plan was to assess the variability in the measurements separately from variability due to age or knee joint health. Following preliminary measures of body composition, height, and weight, an electret and MEMS microphone were both positioned at the lateral and medial sides of the subject's patella targeting the patellofemoral joint while two piezoelectric film sensors were placed on the skin just proximal and distal to the patella. Each sensor was attached using Kinesio Tex tape. In addition to the tape, a thin piece of silicone (5-mm thick) was placed over the piezoelectric film to reduce the interface noise of the tape rubbing against the film. Finally, two wireless inertial measurement units (IMUs) (MTW-38A70G20, Xsens, Enschede, The Netherlands), which contained three-axis accelerometer, gyroscope, and magnetometer as well as built-in sensor fusion outputs, were positioned on the lateral sides of the thigh and shank. These sensor placements are displayed in Fig. 2(a).

While wearing these sensors, each subject completed two exercises: 1) seated, unloaded knee flexion/extension and 2) sit-to-stand. For each exercise, the subject repeated the motion five times while the microphone and IMU outputs were recorded in a quiet room [see Fig. 2(b)]. The signals from the piezoelectric and MEMS microphones were passed through custom circuits and then collected at 50 kHz (16 bits/sample) using Biopac data acquisition hardware (Biopac Systems Inc, Goleta, CA, USA) while the signals from the electret microphones were sampled at 44.1 kHz (16 bits/sample) using a Zoom H6 recorder (Zoom Corp., Tokyo, Japan). The IMU signals were acquired at 50 Hz (16 bits/sample) using their device-specific software suite (MT Manager, Xsens, Enschede, The Netherlands) synched with the Biopac system. Apart from the electret microphone signals, which were stored on an SD card (SanDisk, Milpitas, CA, USA) via the Zoom recorder, all signals were recorded on a laptop. The data were then processed using MATLAB (The Mathworks, Natick, MA, USA).

E. Joint Sound Processing

The signal processing consisted of 1) calculation of knee joint angle and contextualization of the joint sounds with joint angle, 2) identification of significant high frequency acoustic emissions or clicks, and 3) statistical analysis to quantify the consistency of occurrence of the main clicks with respect to joint angle.

First, the knee joint angle was calculated using the methods described in [28], which leverage the sensor fusion outputs of three-axis accelerometer, gyroscope, and magnetometer provided by Xsens, namely the rotation matrix (i.e., Direction Cosine Matrix), and the kinematic constraints of a hinge joint to provide angle data. This method allowed for arbitrary sensor placement and orientation on each segment of the joint (i.e., thigh and shank), eliminating the need for precise calibration techniques and measures [28]. However, this method is poten-

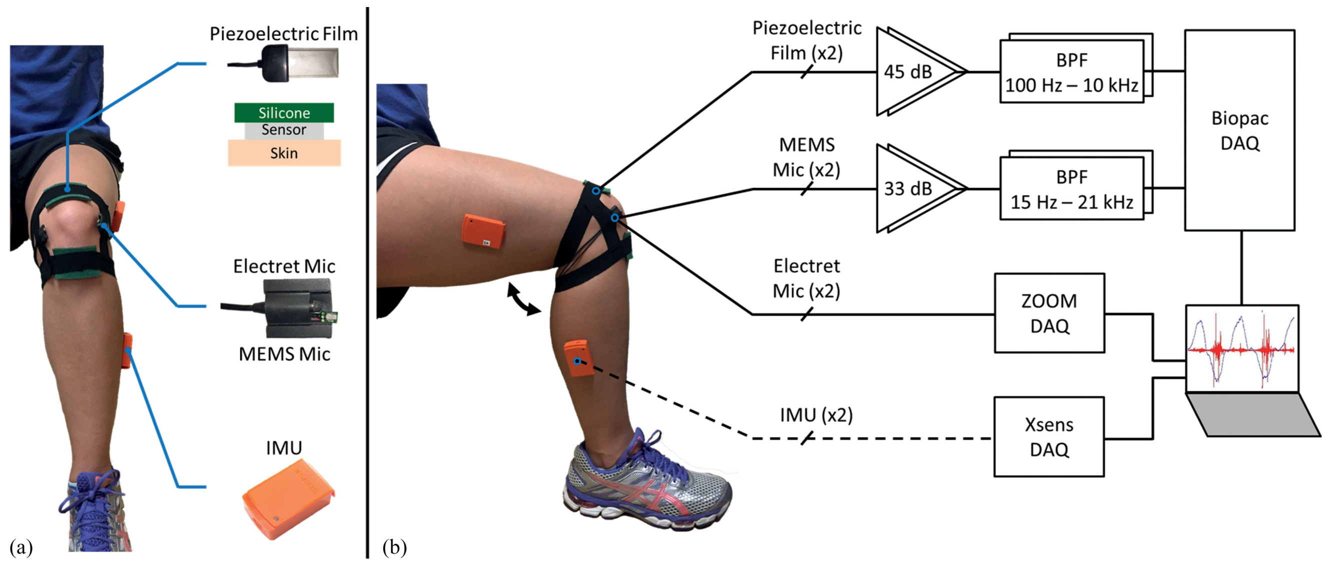

Fig. 2. Sensor placement and measurement block diagram. (a) Eight sensors were used during human subject testing. Two IMUs were placed laterally on the thigh and shank. Piezoelectric film sensors were placed directly proximal and distal of the patella. The air microphones (MEMS and electret) were attached on the lateral and medial sides of the patella. (b) Block diagram of the data collection hardware used during human subject studies.

tially susceptible to error, due to deviations from a true hinge joint as a result of skin and motion artifacts [28]. Nevertheless, since we analyzed the cycles of repetitive motions against one another, this error was common to each cycle and, thus, did not present in our results. Finally, the signal was normalized between 0° and 90° such that subjects could be compared against one another with respect to location within each subject's range of motion.

Once knee joint angle was calculated, the phases (flexion or extension) of each cycle were determined [see Fig. 3(b)]. First, the inflection points of the signal were found. The beginning and ending of each phase were identified by finding pairs of inflection points. These points were then used to segment the microphone signals, contextualizing the data by type of angular motion.

Next, significant acoustic emissions were identified. The most distinct audio signals that were detected by the air microphones were the high amplitude, short duration clicks [see Fig. 3(a)]. By observing the signal's frequency content (i.e., short-time Fourier transform), these clicks were broadband with frequencies as high as 20 kHz. This unfiltered signal contained two main sources of noise; ambient noise ranged in frequency up to 7 kHz while interface noise appeared as baseline movement with components up to 1.5 kHz. The first step of this identification stage was to preprocess the signal such that the clicks became more prominent and any interface and/or ambient noise were mostly cancelled. To this extent, the air microphone signals were filtered with a bandpass filter spanning 7–16 kHz. As seen from Fig. 3(a), the filtered signal ($x[n]$) lacks the original baseline movement, and the clicks are more distinct from other artifacts in the signal.

After this preprocessing step was complete, a modified envelope detection algorithm was implemented. A 1024-bin spectrogram of the signal ($X[n, m]$) was calculated with a window size of 100 samples (i.e., 2 ms) and 90% overlap. The amplitude of the signal was calculated by summing the logarithmic amplitude of the spectrogram across the frequency bins as follows:

$$A[n] = \sum_j 20 \log |X[n, j]|. \tag{1}$$

A moving average and standard deviation ($\mu[n]$ and $\sigma[n]$) of $A[n]$ using a window size of 1000 samples was calculated. $A[n]$ was then thresholded such that

$$T[n] = \begin{cases} A[n], & A[n] > \mu[n] + \alpha \cdot \sigma[n] \\ 0, & \text{otherwise} \end{cases} \tag{2}$$

where $T[n]$ is the thresholded amplitude signal and α is a constant control coefficient, which was selected as 3.3 by inspection.

Next, the peaks of $T[n]$ were detected by standard peak detection techniques. The peaks that resulted from the same click (i.e., resonances of the initial click, which are specified as peaks within 150 samples of each other) were eliminated, resulting in the raw click locations vector $p_r = [p_{r1}, p_{r2}, \ldots, p_{rL}]$. The raw click locations p_r were refined such that each click location corresponded to the point on the original filtered signal where the click achieved its maximum amplitude, positive or negative. The refined click locations matrix $p = [p_1, p_2, \ldots, p_L]$ gave the final detected click locations. An example of these detected clicks is shown in Fig. 3(b).

Once the clicks were identified, we analyzed the consistency of these acoustic emissions. Fig. 3(c) provides a visualization of consistent acoustic emission during repetitive motion. For each cycle of a particular exercise (i.e., flexion or extension), the three clicks with the largest amplitudes and their corresponding angular locations were determined. Each combination of the clicks across cycles (i.e., selection of one of the three clicks from each cycle) was found. The combination with the smallest standard deviation for angular location yielded the most consistently occurring major acoustic event. The mean and standard deviation of these locations were calculated.

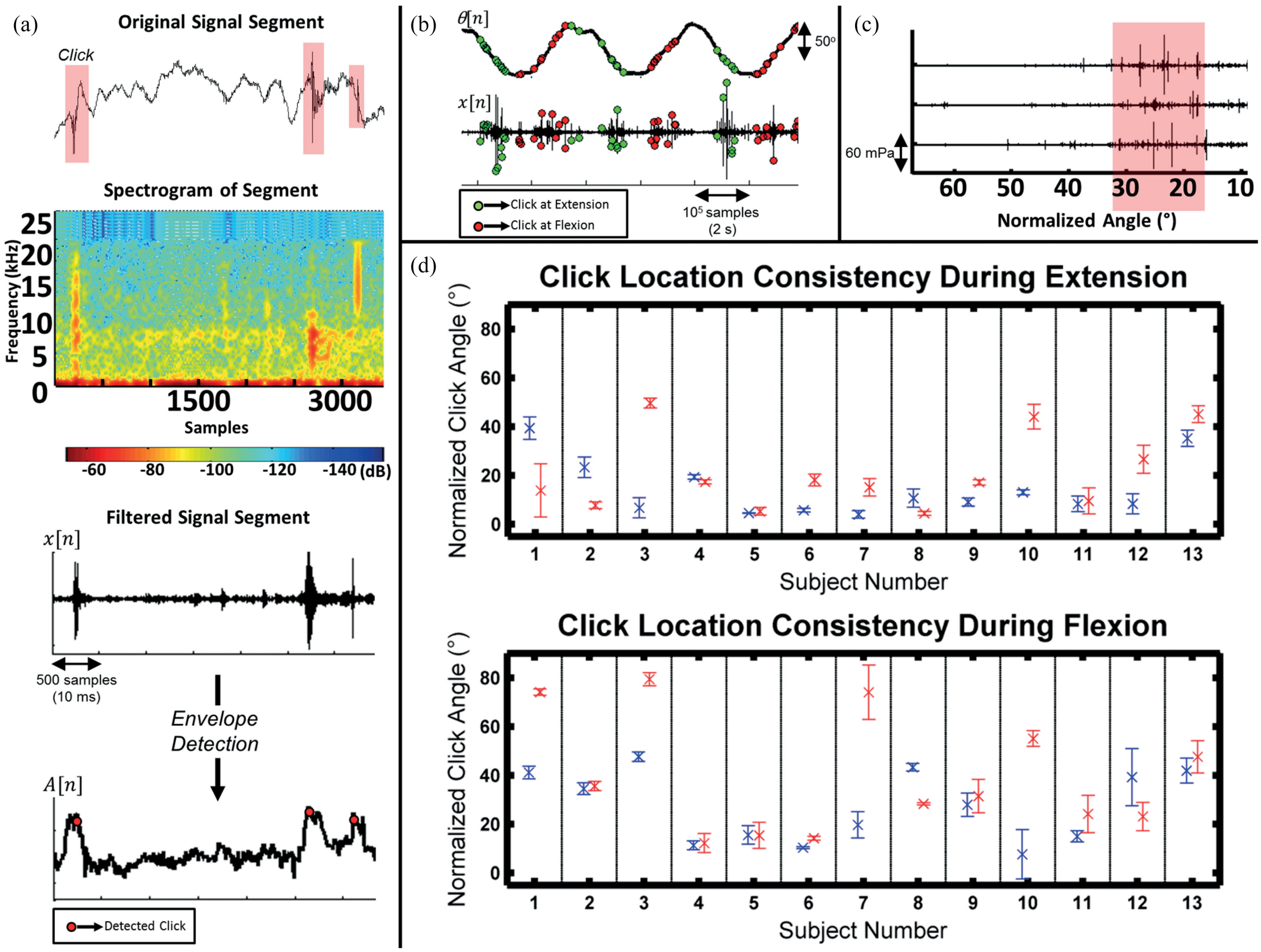

Fig. 3. Joint sound processing of recordings taken with an electret microphone positioned at the lateral side of the patella (a–c) and results (d). (a) An example 3000 sample (60 ms) joint sound recording window showing three distinct high-amplitude, short-duration acoustic emissions. The original signal contains ambient noise, which presents as broadband signals up to 7 kHz, and interface noise, which appears as baseline movement. These components are clearly visualized in the spectrogram of the original signal. To remove the majority of the noise, the signal is bandpass filtered at 7–16 kHz, resulting in the filtered signal $x[n]$. The envelope of this signal is found, yielding $A[n]$. Using a thresholding technique based on the moving average, the significant peaks of $A[n]$ are found, roughly corresponding to the clicks of the original signal. These are later refined to match the true locations of the clicks found in the original signal (i.e., such that the locations correspond to where the clicks achieve their maximum amplitudes, positive or negative, in the original signal). (b) Final result of the click detection algorithm, which displays the identified clicks for three cycles of flexion/extension. (c) Three extension cycles with artificial offsets. These qualitatively show that the main acoustic event of each cycle occurs at similar angular locations. (d) The final results of click location consistency for five repetitions of flexion/extension for 13 subjects on left (blue) and right (red) legs. Across subjects, the standard deviation for click location is small, supporting observations of consistent angular location cycle-to-cycle. Additionally, the mean locations of these clicks are consistent between left and right legs for most of the subjects.

Given these mean locations, three methods were used to analyze the data. For the first two methods, test–retest reliability was estimated using the ICC. We organized the data into "motions" and "repetitions." There were 52 "motions," one for each human subject and exercise combination (e.g., subject 1's extension data represented one "motion"). The "repetitions" consisted of the five click locations (one per cycle) from the selected combination. This dataset will be referred to as the test–retest dataset. Given this dataset, two ICC values were calculated using one-way random single [i.e., ICC (1, 1)] and average measure [i.e., ICC (1, k)] models to show the reliability of a single cycles measure and mean of the five cycles' measures. Additionally, the 95% confidence intervals (CI) for these two ICC values were determined. The last method for analyzing the data was a paired t-test, which was used to assess whether there were significant differences between the mean click locations for left and right legs.

III. Results and Discussion

A. Microphone Comparison

In evaluating our microphone selection, we considered many different parameters. First, we compared the similarity of the signals measured by the electret and MEMS microphones. We also determined the quality of these microphones by evaluating the quality of their sensing capabilities in terms of SNIR. Moreover, when investigating interface issues for the air microphones, we examined the effect that the sensor-to-knee distance had on the signal acquired. Finally, we researched the quality of the contact microphone.

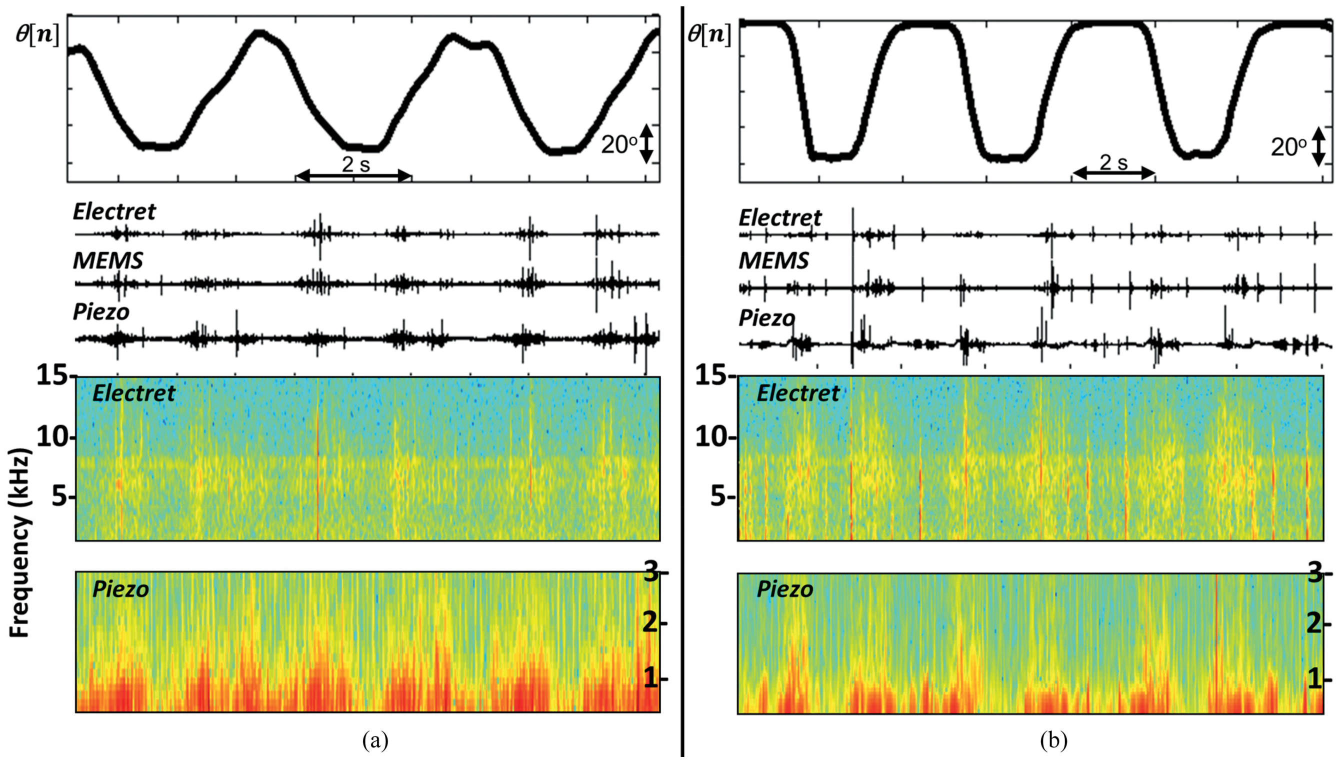

Fig. 4. Joint sounds simultaneously sensed by electret, MEMS, and piezoelectric film microphones during three repetitions of (a) flexion/extension and (b) sit-to-stand exercises. For both parts (a) and (b), the top plot displays the joint angle ($\theta\,[n]$). The middle and bottom graphs show the time and frequency domain signals from the various microphones. The acoustic signatures of the electret and MEMS microphones exhibit similar characteristics.

As shown in Fig. 4(a) and (b), the electret and MEMS microphones, measuring frequencies as high as 20 kHz, performed similarly in detecting joint sounds, which were acquired from a subject performing flexion/extension and sit-to-stand exercises, respectively. This was confirmed by computing the information radius between the normalized histograms of signals captured by these two microphones, which yielded a value of 0.0025. This value shows a high similarity between these two types of microphones since the information radius ranges from 0 for identical distributions to 2 for maximally different distributions [29]. This shows that the more cost-effective MEMS microphones are a viable substitute for the more expensive electret microphones. This is an important result when designing deployable systems.

As predicted, the signal recorded by the air microphones included noise and interface components in addition to the desired joint sounds; both ambient background interference and interface noise caused by the rubbing of athletic tape, which was used to hold the sensors in place, were sensed by the microphones. The SNIR was 11.7 dB for the electret microphone and 12.4 dB for the MEMS microphone. To minimize issues with noise during initial experiments, measurements were taken in a quiet room; however, this will need to be addressed for implementation of a deployable, wearable system, especially given the fact that many background noises, such as speech and sounds due to ambulatory motion, will reside in-band with the joint sounds.

One important observation made during proof-of-concept experiments showed that the air microphones did not need to be directly located at the skin surface to detect airborne joint sounds. As shown in Fig. 5, the sounds obtained from an electret microphone placed on the skin and one located 5 cm off the skin captured similar acoustic signals in both morphology and timing ($\sim 4.5\times$ smaller amplitude). This is an important observation

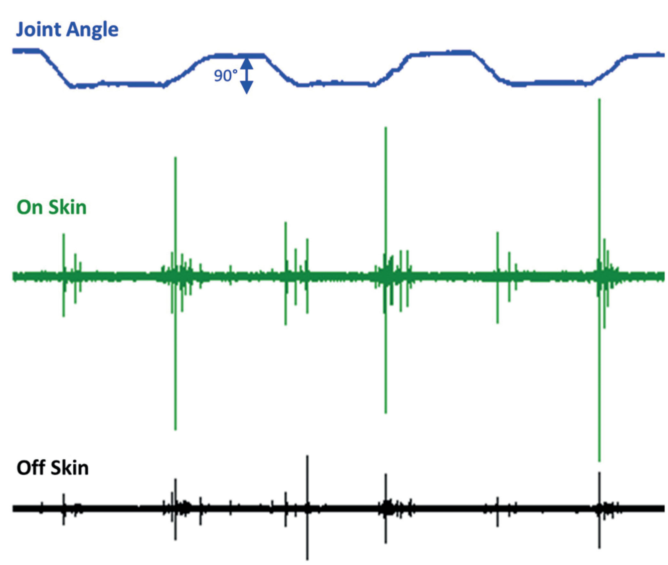

Fig. 5. Joint sounds measured on the skin and 5 cm off the skin during flexion/extension exercises. Though the off-skin microphone captured a signal with decreased amplitude, the on- and off-skin measurements showed significant similarities in their acoustic signatures. The main acoustic event of each signal occurred at similar locations.

because it suggests that the air microphones will be able to record joint sounds in a wearable device where direct contact with the skin may not be constant. However, it will be important to consider this distance when analyzing the captured signals, particularly when the analysis depends on the amplitude of the signal. In this sense, maintaining a fixed distance between the microphone and skin, especially for use in longitudinal analysis, will be required. Furthermore, placing the microphone off of the skin introduces increased potential for noise; the microphone may have a greater opportunity to strike or rub against the skin. Additionally, changing the distance between the microphone and skin will change the microphone's sensitivity in sensing these sounds. These issues must be addressed in the design of a wearable system.

The piezoelectric film measured signals up to approximately 3 kHz as seen from the spectrograms of the signals acquired shown in Fig. 4. While the piezoelectric film had the advantage of not detecting background noise, it acquired significantly more interface noise—8.4 dB SNIR—due to the sensor rubbing on the skin and the athletic tape rubbing on both the skin and the sensor. This interface noise had frequency components up to 1.5 kHz and was thus in-band.

During early pilot data collections, the piezoelectric film was attached to the skin using only Kinesio Tex tape. However, this method proved to be very susceptible to interface noise; as the knee extends and flexes, the tape, though stretchable, deformed the film which obscured the low frequency and low amplitude signatures. Furthermore, though acceptable for collecting pilot data, tape proves to be undesirable for long-term monitoring. To mitigate this issue, a piece of silicone was placed above the piezoelectric film. Because silicone has similar compliant mechanical properties to skin and subcutaneous tissue, the joint sounds received did not experience dampening, and the silicone surface provided a suitable surface to stick the tape. Though this method did not completely eliminate interface noise—the sensor still experienced some movement along the skin—it did help to reduce the recorded noise.

Accordingly, while using piezoelectric film or other contact microphones is desired to capture the vibration signal, which represents the majority of the acoustical energy generated, implementation presents practical issues. The piezoelectric film was significantly affected by interface noise; a smaller portion of the signal bandwidth was corrupted by interface noise for the air microphones compared to contact microphones. Furthermore, contact microphones did not pick up higher frequency vibrations as distinctly as air microphones. For these reasons, at this time, we suggest the use of air microphones for wearable joint sound measurements.

B. Joint Sound Consistency

Fig. 3(d) summarizes the results for mean angular click location for the left and right legs of 13 seated subjects performing five repetitions of knee flexion/extension. Two important findings resulted from this data: 1) significant acoustic events are repeatable during single trial measures and 2) left and right legs produce similar sounds.

First, two ICC values were found for the test–retest dataset. An ICC (1,1) value of 0.94 with a 95% CI of 0.92–0.97 and an ICC (1, k) value of 0.99 with a 95% CI of 0. 98–0.99 were calculated. Since the ICC values were greater than 0.7, these values showed that the main acoustic emission per cycle of activity were consistent within a single trial of monitoring for both single and average measure reliability [30]. Given that audible joint sounds have not been extensively explored, this was an important finding, demonstrating that airborne signals emit a stable pattern with repeated movement in a healthy hinge joint.

Second, the difference between legs for each exercise suggested that a healthy subject's knees produce similar joint sounds; the difference between left and right legs were not significant at the $p < 0.05$ level. While as a group, there were no significant differences between the left and right legs, some subjects could be grouped as having relatively no difference between right limb and left limb click locations whereas others had notable differences between right and left suggesting the potential for defining clinically relevant "signature traits." Such variations in click location could represent useful knee joint health biomarkers.

Though these results are promising, there are some limitations to our current system and analysis. First, with regard to the IMUs, sensor positioning, drift, and motion artifacts can all contribute to flexion angle calculations that differ from the true joint angle. Techniques will need to be employed to minimize these errors, especially when considering their application in a system which measures longitudinal data. For example, some errors could be minimized by ensuring more rigid sensor positioning [28] and leveraging the joint's kinematic constraints directly into the calculation of joint angle [31] to reduce the effect of drift [32]. Second, the effect of lubrication (e.g., diminished boundary lubrication after an injury [33]) and differing structural components (e.g., damaged ligaments [34], etc.) on acoustic emissions has not been sufficiently studied. These variables may introduce "error" when calculating click location consistency for repeated cycles and measuring differences between legs. In this sense, these isolated, one-time measurements may not prove to be as useful as compared to longitudinal analysis for the same subject over time. Future work is required to determine the efficacy of these observations as clinically relevant data.

These quantitative findings in terms of measurement consistency will form the foundation for understanding the significance of changes in joint sound signatures associated with injury, as well as changes in such signatures during rehabilitation. Moreover, while longitudinal studies will be important toward understanding injury recovery, this study and its focus on robust implementation in a wearable platform also presents opportunities for exploring day-to-day and within-day changes of joint acoustics.

IV. Conclusion and Future Work

This paper describes the measurement and analysis of acoustic emissions from the knee joint during loaded and unloaded activities. We demonstrated, quantitatively, that major acoustic events occur at consistent joint angles during repetitive motions for healthy subjects. Furthermore, we observed that these locations are similar between left and right legs for most subjects. Whether asymmetry between right and left knee acoustic emissions is related to risk factors for injury or other training-related variables remains to be clarified. Importantly, these findings showed that joint sound measurements from air microphones are repeatable with sensing technology that can be implemented in an inexpensive, wearable form factor. While extensive analysis of the piezoelectric film was not conducted in this paper due to corruption of the signal with interface noise, we believe its use in a wearable device holds promise based on our preliminary findings showing that packaging techniques have a large influence on the signal recorded.

Future work will include mitigating background and interface noise for both the air and contact microphones. In particular, the focus should be on the packaging of these sensors into a wearable wrap or sleeve enabling high-quality signal measurements during at-home, long-term monitoring. Additionally, existing algorithms should be refined and new processing techniques developed to detect clinically-relevant acoustic signatures. Given that therapists and clinicians look at sounds, swelling, structural stability, and range of motion, researchers should investigate methods for quantifying these joint health biomarkers unobtrusively and accurately; namely, they should determine which acoustic signatures encapsulate these biomarkers. Furthermore, exploration of these biomarkers as they relate to specific diseases and injuries (e.g., osteoarthritis, anterior cruciate ligament tear, meniscal tear, etc.) should be considered. Finally, longitudinal studies on injured subjects will allow determination and validation of specifics acoustic emission features (e.g., consistent angular location) that provide valuable joint health information during rehabilitation following an acute injury.

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