Non-Invasive Assessment of Cartilage Damage of the Human Knee Using Acoustic Emission Monitoring: A Pilot Cadaver Study

Liudmila Khokhlova, *Graduate Student Member, IEEE*, Dimitrios-Sokratis Komaris, Nikolaos Davarinos, Karuppiah Mahalingam, Brendan O'Flynn, *Senior Member, IEEE*, and Salvatore Tedesco, *Member, IEEE*

***Abstract*—*Objective:* Knee osteoarthritis is currently one of the top causes of disability in older population, a rate that will only increase in the future due to an aging population and the prevalence of obesity. However, objective assessment of treatment outcomes and remote evaluation are still in need of further development. Acoustic emission (AE) monitoring in knee diagnostics has been successfully adopted in the past; however, a wide discrepancy among the adopted AE techniques and analyses exists. This pilot study determined the most suitable metrics to differentiate progressive cartilage damage and the optimal frequency range and placement of AE sensors. *Methods:* Knee AEs were recorded in the 100–450 kHz and 15–200kH frequency ranges from a cadaver specimen in knee flexion/extension. Four stages of artificially inflicted cartilage damage and two sensor positions were investigated. *Results:* AE events in the lower frequency range and the following parameters provided better distinction between intact and damaged knee: hit amplitude, signal strength, and absolute energy. The medial condyle area of the knee was less prone to artefacts and unsystematic noise. Multiple reopenings of the knee compartment in the process of introducing the damage negatively affected the quality of the measurements. *Conclusion:* Results may improve AE recording techniques in future cadaveric and clinical studies. *Significance:* This was the first study to evaluate progressive cartilage damage using AEs in a cadaver specimen. The findings of this study encourage further investigation of joint AE monitoring techniques.**

***Index Terms*—Acoustic emission, joint sound, knee health, osteoarthritis, thinning of the articular cartilage.**

Manuscript received 24 June 2022; revised 14 December 2022 and 16 March 2023; accepted 25 March 2023. Date of publication 30 March 2023; date of current version 30 August 2023. This work was supported in part by Science Foundation Ireland (SFI) under Grant 12/RC/2289-P2 (INSIGHT), which was co-funded by European Regional Development Fund (ERDF), and in part by SFI under Grant 13/RC/2077 for CONNECT and Grant 16/RC/3918 for CONFIRM, which was also co-funded by ERDF. *(Corresponding author: Liudmila Khokhlova.)*

Liudmila Khokhlova is with the Insight Centre for Data Analytics, Tyndall National Institute, University College Cork, T12 K8AF Cork, Ireland (e-mail: liudmila.khokhlova@tyndall.ie).

Dimitrios-Sokratis Komaris was with the Insight Centre for Data Analytics, Tyndall National Institute, University College Cork, Ireland. He is now with the Aston University, U.K.

Nikolaos Davarinos and Karuppiah Mahalingam are with the Mater Private Network, Orthopaedic & Spine Centre, Ireland.

Brendan O'Flynn and Salvatore Tedesco are with the Insight Centre for Data Analytics, Tyndall National Institute, University College Cork, Ireland.

This article has supplementary downloadable material available at https://doi.org/10.1109/TBME.2023.3263388, provided by the authors.

Digital Object Identifier 10.1109/TBME.2023.3263388

I. Introduction

THE knee is one of the largest, most complicated and heavily loaded joints in the human body. The knee constantly sustains high loads, which can be multiple times greater than a person's bodyweight, during daily routine and sporting activities, such as walking and running. This leads to a high prevalence of injuries and pathologies, for instance knee osteoarthritis (OA), which is characterised by the deterioration of the joint's cartilage and underlying bone. According to the 2019 Global Burden of Disease report [1], knee OA is considered to be one of the top ten causes leading to disability in the European region for the population aged above 70 years and in the top fifteen for population between 50 and 70. The prevalence of knee OA has grown globally since 1990 and this trend is expected to continue due to the population ageing and the increasing levels of obesity [2], [3].

Currently, the clinical assessment of the joint's condition mostly relies on imaging, patient-reported symptoms, functional tests and physical evaluations (e.g., mobility evaluation). However, the early diagnostic of OA is still challenging. The currently used imaging technologies display some limitations: radiography methods (X-rays and CT scans) are not sensitive to minor changes over time and do not adequately visualize soft tissue; ultrasound imaging shows relatively poor resolution and has limited observation area due to bone obstructions; whereas MRI provides excellent anatomic resolution, it remains costly and long imaging time is needed; finally, nuclear technologies are expensive and involve additional exposure to radiation [4]. At the same time, patient-reported outcomes are relatively subjective and not sufficiently sensitive [5], and functional tests may only represent a specific physical function [6]. Moreover, due to the recent global focus on telemedicine and remote assessment [7], [8], the need for diagnostic tools that are not tied to a clinical site is evident. Therefore, digital technologies that can provide interactive remote orthopedic examinations are expected to remain the focus of future research [9]. With the current advances in the area of electronics and wearable technologies [10], alternative solutions, such as inertial sensors [11], [12], [13] and acoustic

This work is licensed under a Creative Commons Attribution 4.0 License. For more information, see https://creativecommons.org/licenses/by/4.0/

emission (AE) monitoring [14], [15], [16], have the potential to be used as part of joint condition assessments.

Among the methods that are potentially suitable for telemedicine applications for joint health and implant condition assessment, AE monitoring remains one of the least investigated. This technique is widely used in non-destructive testing to detect the occurrence and location of defects in structures of various kinds by sensing acoustic waves that occur during deformation. The method is also used to monitor phase transformation, friction and corrosion of materials. Signal processing in AE monitoring usually involves acoustic event (hit) detection. Hit detection is typically defined by the following parameters: signal amplitude threshold (usually expressed in dB) for which a hit is detected; hit definition time (HDT), which specifies the maximum time between two consecutive threshold crossings; hit lockout time (HLT), defined by the time that must pass after a hit has been detected and before the next one occurs and can be detected again; and peak definition time (PDT) which is determined by the time from hit detection to the hit's peak [17]. Several parameters of the detected hits can be measured to gain a further understanding of the investigated process or defect. Such parameters may include, but are not limited to, the number of hits detected in a specified time, hit durations, the average time between hit detection and its peak (rise time), and the average frequency of the signal in a hit.

AE monitoring in orthopaedics was initially used to investigate structural and biomechanical properties of bones [18]; however, in recent years, non-invasive joint diagnostics have received a renewed focus in the field. A scoping review was carried out by the authors [19], showing that AE monitoring is a promising technique in the diagnostics of joint pathologies, such as age-related deterioration [20], [21], OA [22], [23], [24], and past injuries [15]. A wide variety of methods is currently suggested for joint AE acquisition and processing, including many diverse approaches in AE sensor placement and fixation, different signal frequency ranges and lower limb movements. However, further research is needed to establish optimal methods of a robust recording procedure, suggest a unified methodology, enable a useful exchange of data, and ensure successful translation of the method to wider research practice and, in future, to clinical applications.

In terms of the data acquisition approaches in the existing literature, the frequency range of the employed sensors varies greatly within studies on joint sounds. For instance, among the reviewed papers in [19], the sensors' acquisition frequency varied widely from as low as 1kHz to higher ranges of up to 500 kHz. Moreover, a lower frequency range below 1kHz was used in vibroarthography [25]. It has been also shown that the highest energy joint sounds are present in frequency ranges below 25kHz [26], however, positive results in differentiating between OA and healthy knees were also achieved with higher frequencies of up to 400 kHz [24], [27], [28]. This could potentially indicate that while amplitude of AE events is smaller in the high frequency range, the recordings in this range might be less sensitive to noise caused by motion artefacts, and need to be considered as potentially useful in practical applications.

With regards to the influence of motion artefacts, ensuring optimal sensor position and attachment, is an essential requirement in obtaining AE recordings of a good quality. It has been previously shown that the appropriate placement of an AE sensor on the joint plays a significant role in achieving repeatable results and an acceptable signal-to-noise ratio [29], [30]. However, to date, only a small number of studies have been conducted to specifically address the questions of optimal sensor placement and fixation, as well as the potential adverse effects of improper fixation. For instance, the work by Ozmen et al. [31] discusses mounting methods to improve accelerometer sensing performance in the 50 Hz-10 kHz frequency band for sound recordings of the knee. In our previous work [32], [33], we presented a procedure to mitigate motion artefacts and improve the repeatability of AE recordings in the broad frequency range of 20-500 kHz. Finally, different positions for the placement of AE sensors were considered for OA knees [21] and wrist joints of healthy volunteers [30]. Though certain AE sensor positions and additional measures were suggested to minimize influence of motion artefacts and improve signal to noise ratio, to the best of the authors' knowledge, the sensor positioning was never considered with regard to the location of the cartilage damage within the knee compartment and potential attenuation of the AE signal.

Finally, in respect to studies where precise damage was inflicted on the cadaveric specimens, Whittingslow et al. [34] investigated the nature of AEs after simulated surgery, with artificially induced meniscus damage and saline injections to imitate tissue swelling. The authors focused on the relationship between internal contact of the articulating structures within the knee and the produced AEs. The reported results show that a meniscus tear increases the number and amplitude of the AE peaks as compared to the baseline recording, while a large increase in the AEs was observed during minimal inter-joint distances. In this regard, the use of cadaveric specimens allows the recording of AEs in a highly controlled manner, and ensures the infliction of precise damage in a specific area. Although the authors in [34] simulated both meniscus damage and soft tissue swelling, cartilage damage that is inextricably related to knee OA has yet to be considered in cadaveric models.

In view of the above, the aim of this study is to determine the optimal frequency range and sensor placement of an AE sensor, along with the most suitable metrics to differentiate progressive degrees of cartilage damage on different articulating surfaces of the knee joint. The present pilot cadaveric study was conducted using two AE sensors with dissimilar frequency responses and a high sampling rate, allowing for the recoding of the different frequency components of the joint AE and their comparison under controlled conditions. Such conditions, i.e., same specimen, cartilage damage degree and location, joint load and movement, to the best of authors' knowledge, have not been implemented before. Moreover, alternating placements of the sensors on the lateral and medial condyles of the knee were included to compare the sensors' response and joint AE characteristics with regards to the precise location of the damage, which also have not been explored in detail before.

We hypothesized that the AE sensor placed closer to the damaged compartment would detect higher number of AE events as well as recording AEs in the lower frequency range, resulting in better differentiation between the tested conditions of progressive damage.

II. Materials and Methods

A. Specimen Preparation

The present study was conducted on a fresh-frozen human cadaveric lower limb. The specimen was procured from and tested at the ASSERT Centre, College of Medicine and Health, University College Cork. All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and/ or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards. Consent for publication of research was obtained at the time of body donation. The donation and dissection of human bodies, for medical education, research and training is covered by legislation governing the practice of anatomy in the Republic of Ireland (Anatomy Act 1832, Medical Practitioners Act 2007).

The hemi-pelvis to toe-tip specimen was thawed at room temperature for 36 hours prior to testing. The donor (male, 32 years of age) had no known major knee injuries or other musculoskeletal disorders. The specimen was fixed on a surgical table with heavy-duty straps in a manner that allowed its unobstructed motion, and was then preconditioned with manual flexion/extension movements for five minutes. No presence of excessive subcutaneous adipose tissue or cartilage abnormalities were observed upon the opening of the knee compartment.

B. AE Sensors and Equipment Setup

The USB AE Node (Physical Acoustics) monitoring system and the corresponding AEwin software (Mistras, Physical Acoustics) were used to record AE events from the knee joint. Two AE sensors, the PK15I and the PK3I, with embedded preamplifiers were used to capture AEs, using AE node integrated filters with 100-450 kHz and 15-200kH ranges, respectively. Sensors were selected from commercially available, ready-to-use, low-noise, small alternatives in the medium frequency band, that have been previously used successfully for the recording of joint acoustic emissions (AEs) [19]. Employing two sensors also allowed the recording of signals in relatively broad frequency ranges. Hit definition parameters were pre-set to the following values: PDT $= 200$ μs, HDT $= 800\mu$s, HLT $=$ 1000 μs, based on the results of Shark et al. [35] and according to the previously used frequency range in the literature [19]. Registration threshold was set at 24 dB for the PK15I - or henceforth referred to as high frequency (HF) sensor, and 32 dB for the PK3I - denoted here as low frequency sensor (LF). The sensors weigh 51 gm, and have a height of 72mm and a diameter of 27mm. A previously investigated method of sensor attachment resistant to motion artefacts [32] was used in this study with some modifications. In particular, ethylene-vinyl acetate (EVA) foam (100kg/m^3) holders were glued to the skin using cyanoacrylate glue instead of double-sided tape. To avoid accumulation of excess glue on the sides of the holders, a small amount of glue was used and was distributed evenly in a thin layer on the foam's surface. The centre of each holder contained a cylindrical cut-out aperture to hold the AE sensor and an indentation for the connecting cable. The sensor was then inserted into the foam holder until firm contact with the specimen's skin was achieved. The HF sensor was initially placed on the medial tibial condyle area of the knee, while the LF sensor was positioned symmetrically on the lateral side. The sensors' position was chosen according to indications from the literature [21] for sites with minimum muscular and dynamic artefacts. However, placement of the sensors on the apex of the patella was unfeasible due to the position of the incision.

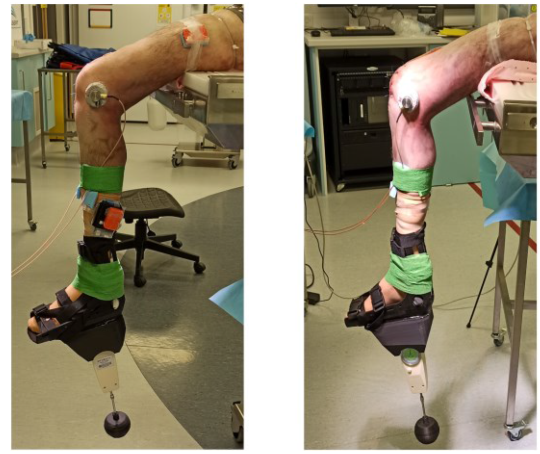

Fig. 1. Testing setup: Lateral side (left) and medial side (right).

To generate AEs, the knee was manually flexed/extended with a moderate load (100 ± 10 N) that was applied indirectly to the joint with the use of an ankle brace and a force gauge (Fig. 1). The force meter provided visual feedback to the operator who maintained and verbally confirmed the stable loading throughout the motion cycles. The ankle brace immobilized the ankle and facilitated the manipulation of the leg in one plane while constraining all ankle movements. All the surfaces of the ankle brace that were in contact with the specimen were isolated with EVA foam (5mm thickness, 45kg/m^3) to avoid potential noise due to friction. Additionally, all connecting cables were isolated with foam and secured with holders on the bench to avoid excessive movement. To track the manipulation of the limb, inertial measurement units (IMUs) were placed on the shank and thigh of the specimen, and were secured with medical tape and elastic straps (Fig. 1, left). The position of the sensors was set to zero in a fully-extended leg position.

C. Data Acquisition

AE events were recorded during 10 flexion/extension cycles at each step. The baseline recording (BL) was obtained from the intact knee using both AE sensors (HF and LF). The knee joint internal surfaces were then exposed following the medial parapatellar approach [36] by two consultant orthopaedic surgeons that routinely perform knee arthroplasties in their practice.

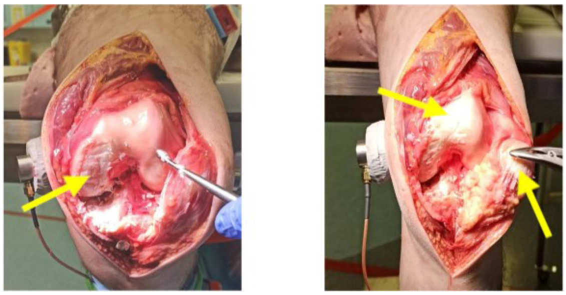

Fig. 2. Cartilage damage: Medial (MED, left) and patellofemoral (MEDPAT, right).

Initially, the femur cartilage surface of the medial compartment (MED) of the knee was scraped with a scalpel and a bone rasp simulating stage III osteoarthritis as per the Kellgren and Lawrence (KL) system (Fig. 2, left).

The incisions were then closed with nylon stitches and subsequently, the knee was manipulated again into another round of flexion/extension. Subsequently, the sensors were swapped by placing the HF sensor on the lateral and the LF on the medial sides into the respective holders, and a third round of knee manipulation was recorded. The sensors were then returned to their original configuration and all further recordings were made with the HF in the medial and the LF in the lateral positions. As the foam holders were glued to the skin, the sensors' positions remained the same with regards to the tissue below during sensor exchange.

To further investigate AE readings during increasing damage, the cartilage was progressively removed on different areas of the joint. The knee compartment was reopened and additional damage was inflicted onto the articulating patellofemoral surface (MEDPAT) (Fig. 2, right). Then, new stitches were put in place and knee movements were recorded in the same manner as above. Afterwards, all measurements were repeated upon inflicting further damage onto both the medial and patellar compartments to simulate grade IV of OA (MEDPAT4Gr). Lastly, the process was repeated after also damaging the lateral compartment (LAT) of the joint followed by the previously described closing of the knee and recording procedure. The block diagram of the experimental process presented in the Fig. 3.

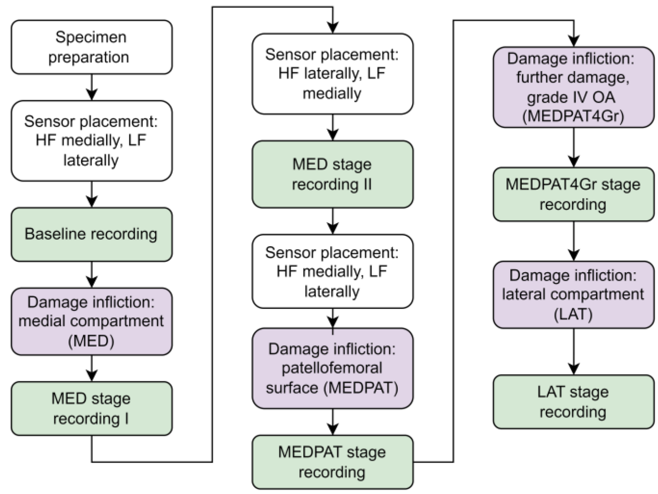

Fig. 3. Block diagram of the experimental process.

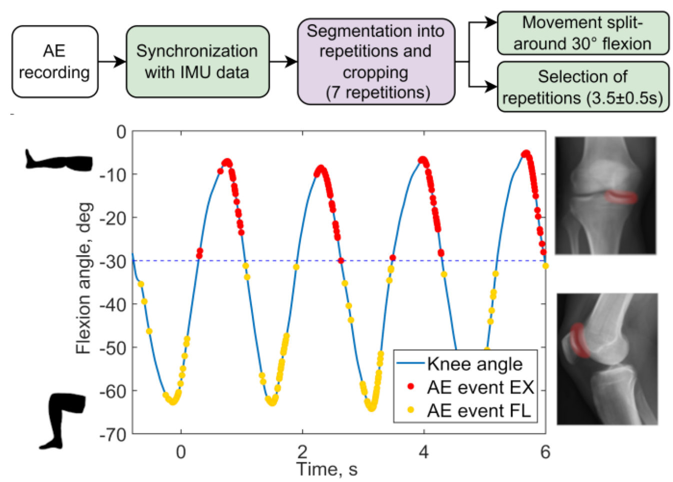

Fig. 4. AE recording processing (top) and segmentation (bottom): Contact areas of the damaged cartilage surfaces in relation to the flexion angle from 30° to full extension (EX) and from 30° to maximum flexion (FL).

D. Signal Processing and Data Analysis

The recorded AE hits and sensor orientation data were analyzed (Fig. 4, top) using MATLAB (Mathworks). Timestamps were used to synchronize the AE recordings with motion data, and the angular movement of the shank was utilized to segment the recorded signals into repetitions. Movement from maximal extension (straight leg) to flexion at around 90 degrees angle and back was considered as one repetition. Seven repetitions out of the recorded ten were used on average, from full extension to maximum knee flexion, by excluding the start and end of the movement due to additional noise and/or compromised execution in several records (e.g., noise from the operator's hand contact during the initiation and termination of the movement and low load applied at the start or end of the trials). To further increase the homogeneity of the included flexion/extension cycles, the analysis was additionally constricted to include only repetitions in the range of 3.5 ± 0.5 s.

Moreover, the knee movement was split (Fig. 4, bottom) into the range less than 30° of flexion (EX), (approximately half range of motion) and over 30° (FL). Limiting the knee angle to range lower than 30° (EX), concentrates the analysis in the part of the movement with maximum contact area in the medial compartment (MED). Likewise, higher flexion angles (FL) results in a wider contact areas with patella (MEDPAT) [37].

Then, the following AE parameters were exported from the AEwin software: rise time (time between detected AE hit start and its peak amplitude in μs); count (number of an AE signal excursions over the threshold in one hit); hit duration (μs); amplitude (dB); counts to peak (number of an AE signal excursions over the threshold between AE hit start and its peak amplitude); signal strength (pV-sec), absolute energy (attoJoules); average

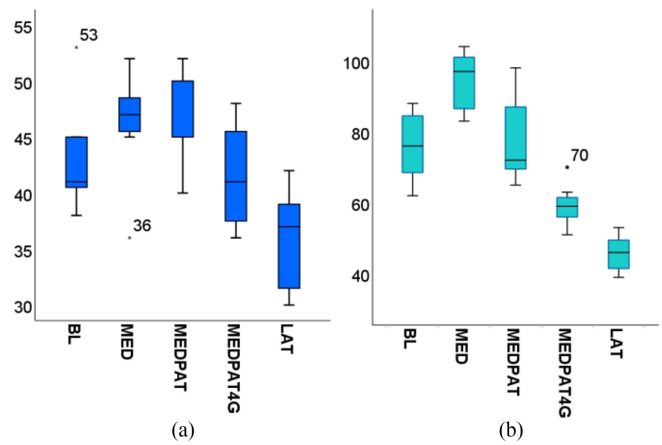

Fig. 5. Box plots of mean number of hits per repetition (for seven repetitions per measurement): (a) HF sensor (b) LF sensor.

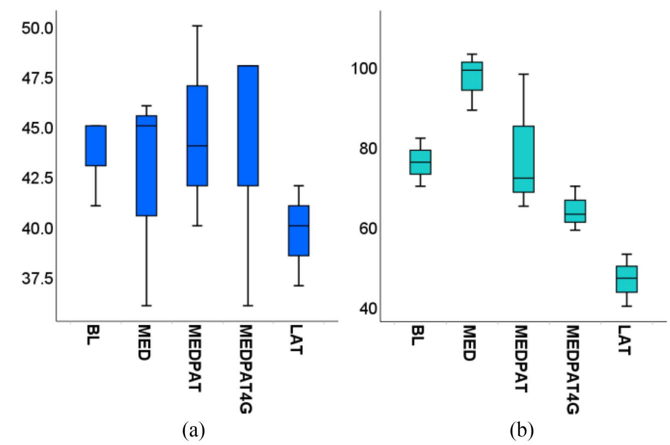

Fig. 6. Box plots of mean number of hits per repetition (for three repetitions with durations of 3.5 $\pm$ 0.5 s per measurement): (a) HF sensor (b) LF sensor.

frequency (average frequency over the entire AE hit in kHz). Means and standards deviations were calculated for all parameters. Detailed description of the parameters can be found in the supporting documentation of the AEwin software [38].

Additionally, amplitude distribution analysis of the acoustic data was conducted using log-linear amplitude/cumulative frequency plots and b-values. These metrics were first proposed by Gutenberg and Richter [39], and are widely used in seismology to quantify the relationship between the magnitude and frequency of a seismic trace. The b-value, is derived from an empirical formula (1), where N is the number of events, M is the magnitude of the event, and a and b are constants, with b representing the negative gradient (slope) of the log-linear acoustic emissions frequency/magnitude plot. The metric had been recently applied to AEs and was successfully used to differentiate knee injury status in athletes [14], and meniscus damage in cadaver specimens [34]. The cumulative number of hits in seven repetitions was used in the analysis. The best line fit was used to calculate b-values.

$$log\ N = a - bM \tag{1}$$

Statistical analysis was performed using IBM SPSS Statistics. Normality was assessed for the AE hit parameters using the Shapiro-Wilk test [40]. If data distribution was presumed normal, independent samples t-tests were used to determine if any statistically significant difference was present between AE parameters for the studied sensor positions and cartilage damage stage. In the case where observations followed non-normal distributions, pairwise independent-sample Kruskal-Wallis tests with Bonferroni corrections were used for comparison. All statistical tests were performed with 95% confidence interval.

III. Results

A. Joint AE Parameters

The means and standard deviations of the detected number of hits per repetition during the manipulation of the intact knee (BL) were equal to 76 $\pm$ 10.12 for the LF and 43.29 $\pm$ 4.99 for the HF sensors (BL, Fig. 5). A larger number of hits, equal to 94.14 $\pm$8.76, was detected for the damaged medial compartment (MED) when compared to the BL recording with the LF sensor; a higher number of hits was also registered with the HF sensor (46.14 $\pm$ 5.01) for the first stage of damage (MED). A slightly higher number of hits (47.43 $\pm$ 4.28) was likewise observed after the medial patellar damage (MEDPAT) with the HF sensor. However, a progressively lower number of hits per repetition was recorded for every recording thereafter (MEDPAT4Gr and LAT for HF sensor, MEDPAT, MEDPAT4Gr and LAT for LF sensor, Fig. 5).

Since longer flexion/extension cycles under constant load would naturally generate more AE events, the analysis was repeated with only the movement cycles that had a duration of 3 to 4 seconds (first 3 repetitions out of 7 per measurement with the specified duration). However, the same trend was observed for LF sensors (Fig. 6(b)), with a higher number of hits per repetition for the first stage (MED, followed by a decline in their numbers in the last measurements. For the HF sensor the decline was less prominent with only the last stage of damage being visibly lower than the baseline.

Due to the process of recording AE signals using a predetermined threshold for hit detection, the gathered data can be skewed. The Shapiro-Wilk test for all the included AE hit parameters (i.e., rise time, counts, duration, amplitude, counts to peak, signal strength, absolute energy and average frequency) showed that data do not follow normal distributions ($P < .001$). To ensure optimal representation of the non-normally distributed parameters, the median values of the exported AE parameters from both sensors are presented in Table I. Additionally, mean values, interquartile ranges and standard deviations are presented in Tables S1 and S2 of the supporting information for the HF and LF sensors, respectively.

For the LF sensor (Table I), hit rise time and the number of threshold crossings (counts) were significantly different for the MED when compared to all other measurements; however, the rise time did not yield any significant differences for all other comparisons, apart from the BL vs. MEDPAT4Gr. Hit durations, signal strength and absolute energy were significantly different between the BL and the MED and MEDPAT stages, that were also significantly different between each other. Counts to peak

TABLE I
JOINT AE PARAMETERS (FOR 7 REPETITIONS)

Damage stage	Rise time, µs	Count	Duration, µs	Amplitude, dB	Counts to peak	Signal strength, pV-sec	Absolute energy, attoJoules	Average frequency, kHz
HF: median								
BL	70	1	27	24	1	198.25	0.22	68
MED	55	1	18	24	1	155.55	0.22	66
MED PAT	44	1	45	24	1	286.7	0.34	39
MED PAT4 Gr	63	1	63	24	1	436.15	0.52	43
LAT	70	1	31	24	1	152.5	0.14	76
LF: median								
BL	47	5	377	40	2	11656	55.21	20
MED	31	3	177	37	1	4648	19.64	21
MED PAT	34	4	261	38	1	8055	33.45	20
MED PAT4 Gr	33	5	283	38	1	7656	32.42	21
LAT	34.5	4	269.5	38	1	7886	34.02	20

MED (medial) - the cartilage surface damage on the medial compartment, KL III; MEDPAT (medial, patellofemoral) - the cartilage surface damage on the medial compartment plus patellofemoral surface, KL III; MEDPAT4Gr (medial, patellofemoral, grade IV KL) - the cartilage surface damage on the medial compartment plus on the patellofemoral surface, KL IV; LAT (lateral) - the cartilage surface damage on the medial compartment plus on the patellofemoral surface and lateral compartment, KL IV.

were also significantly different between the BL and both the MED and MEDPAT, but no significant difference between those two categories (MED and MEDPAT) was observed. The average frequency distribution did not differ for all measurements. The amplitude of the hits was significantly different for the MED, MEDPAT and MEDPAT4Gr when compared to the BL, while a significant difference was also observed for the MEDPAT vs. MED.

For the HF sensor (Table I), the rise time, threshold crossings (counts), counts to peak, hit duration, average frequency, signal strength and absolute energy were not significantly different across all recordings. The hit amplitude was significantly different for the MED and MEDPAT4Gr conditions, but not significantly different for all other pairwise comparisons. The p-values for all comparisons for both sensors, as well as the pairwise diagrams, can be found in the supporting information: S LF.pdf and S HF.pdf.

B. Alternative Measures: B-Value

The following graphs illustrate the amplitude distribution of the registered hits for the two employed sensors. The line of the best fit illustrates the slope or b-value (Figs. 7 and 8, for the HF and LF sensors, respectively). The hits with an amplitude above 48 dB for the HF and above 66 dB for the LF sensors were filtered, to remove outliers due to motion artefacts and noise.

The obtained b-values for the BL, MED, MEDPAT, MEDPAT4Gr and LAT measurements are 0.13, 0.20, 0.15, 0.17, and 0.14 for the HF sensor, and 0.09, 0.13, 0.13, 0.08, and 0.14 for the LF sensor, respectively. The b-values for progressive damage for the LF sensor overall follows the same trend as hits per repetitions with the first measurement with inflicted damage on the articulating surfaces of the medial compartment (MED) showing the biggest difference from the measurement of intact knee (BL), and the following measurements showing values closer to the BL condition. Notably, for the LF sensor, the intact knee showed a higher number of hits for the majority of the amplitude range. However, for the HF sensor, a distinction can be noticed on the amplitude distribution graph between the intact knee (BL, blue line) and all following measurements in the range of amplitudes between 28 and 38 dB, with more hits being registered for all stages of damage as compared to the baseline.

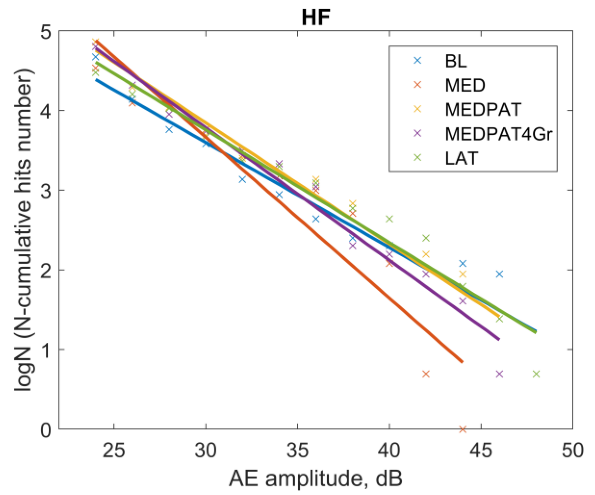

Fig. 7. Cumulative hit occurrence frequency and the amplitude relationship plots for the HF sensor.

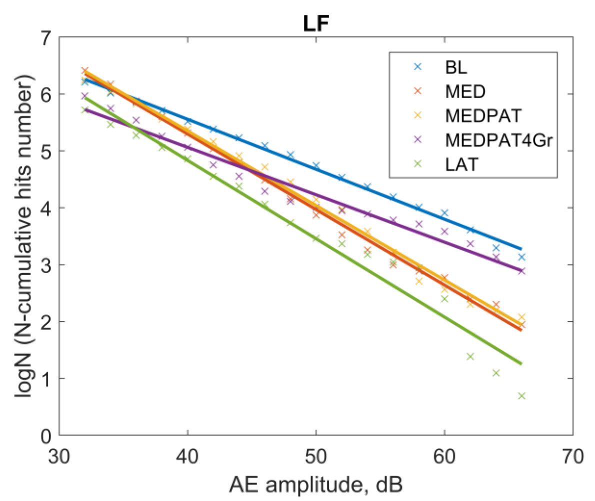

Fig. 8. Cumulative hit occurrence frequency and the amplitude relationship plots for the LF sensor.

C. Sensor Position

The number of hits per repetition that were registered with the HF sensor in the medial position was 46.14 ± 5.01, while 58.86

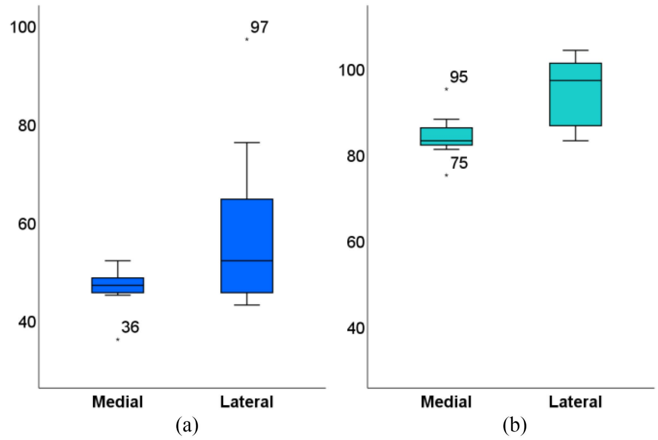

Fig. 9. Box plots of the mean number of hits per repetition (for seven repetitions) in the lateral and medial positions: (a) HF sensor (b) LF sensor.

TABLE II
JOINT AE PARAMETERS (FOR 7 REPETITIONS) IN LATERAL AND MEDIAL POSITION FOR THE CARTILAGE SURFACE DAMAGE ON THE MEDIAL COMPARTMENT, KL III (MED)

Damage stage	Rise time, µs	Count	Duration, µs	Amplitude, dB	Counts to peak	Signal strength, pV-sec	Absolute energy, attoJoules	Average frequency, kHz
HF: median								
M	55	1	18	24	1	155.55	0.22	66
L	25	1	19	24	1	170.80	0.20	114.50
LF: median								
M	44	5	375	40	1	11456	55.35	19
L	31	3	177	37	1	4648	19.64	21

M – medial sensor position, L - lateral sensor position.

± 20.16 hits were registered in the lateral position (Fig. 9(a)). For the LF sensor (Fig. 9(b)) 94.14 ± 8.76 hits per repetition was recorded in the lateral and 84.14 ± 6.18 in the medial positions. Interestingly, more hits were detected when recording laterally of the knee using both the HF and LF sensors, even though damage was inflicted in the medial compartment. The means, standard deviations, medians and interquartile ranges of all computed parameters are presented in Tables S1 and S2 of the supporting information.

For the LF sensor, all included parameters (Table II) were significantly different between the lateral and medial positions. Shorter hit duration, lower values of hit rise time, amplitude, threshold crossings in a hit (count), counts to peak, average frequency and signal strength were observed in the lateral sensor position. The absolute energy of AE hits, however, was lower in the medial position.

For the HF sensor (Table II) rise time was significantly higher in the medial sensor position. Median amplitude values were the same for both positions; however, the distribution was significantly different. Other parameters have not shown any difference between the recordings in different positions. The p-values, as

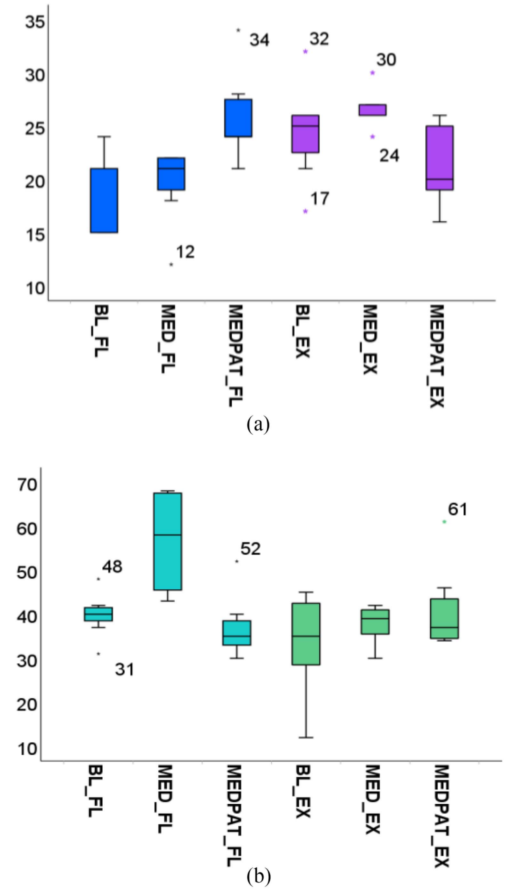

Fig. 10. Box plots of the mean number of hits per repetition (for seven repetitions) in the lateral and medial positions: (a) HF sensor (b) LF sensor.

well as the pairwise comparison diagrams, can be found in the supporting information: S LF.pdf and S HF.pdf.

D. Joint Angular Position

The numbers of hits per repetition that were registered with the HF and LF sensors for the portions of movement from 30° and over (FL), and flexion angles lower than 30° (EX), are presented in Fig. 10. For the HF sensor in the FL segment, bigger number of hits (26.00 ± 4.20) was observed for the MEDPAT in comparison to the BL and MED conditions, as expected, due to the increased contact in the patellofemoral area. However, the same was not observed with the LF sensor, where a lower number of hits was detected (37.15 ± 7.31), despite the increased damage in the articulation area. Interestingly, in the FL segment of the movement, the number of hits was higher for the MED stage (56.43 ± 11.59) that in the measurement for intact knee (BL, 39.86 ± 5.16), indicating increased AE output from the damaged medial compartment. Considering the EX portion of the movement, a higher number of hits was expected after the MED stage, but for both sensors this increase was not substantial (Fig. 10).

For the HF sensor, the duration, average frequency, signal strength and absolute energy did not differ between the FL and

EX portions of the movement, whereas amplitude, rise time of the hits, counts (threshold crossings) and counts to peak were significantly different only for the MEDPAT stage. For the LF sensor, majority of the AE event parameters did not differ between FL and EX portion of movement for BL condition, apart from amplitude and absolute energy. For MED condition, different picture was observed with only three parameters (threshold crossings, counts to peak and amplitude) were not significantly different for movement portions. For MEDPAT stage, only counts to peak and average frequency were not significantly different between FL and EX. The p-values for all comparisons for both sensors, as well as the pairwise diagrams, can be found in the supporting information: S LF.pdf and S HF.pdf.

IV. Discussion

A. Joint AE Parameters

The acquired data in this study partially contradicts the previously reported notion of registering higher numbers of AE events as the knee's cartilage damage progressively increases simulating different grades of OA [24]. The inflicted damage on the articulating surfaces of the medial compartment (MED) when recording with the LF sensor, and the medial compartment (MED) and further patellar damage (MEDPAT) with the HF sensor, were associated with a higher number of AE hits; however, further measurements with increasing damage yielded a smaller number of AE events that ultimately resulted in a lower number of hits per repetition than the baseline measurement (Fig. 5). This was particularly evident with the LF sensor, while a similar behaviour was observed with the recordings of the HF sensor for seven repetitions, but not when considering only repetitions with a specific duration, where only the last stage returned less AE events than the BL (Fig. 6).

All the investigated AE hit parameters (Table I) followed the same trend for the LF sensor, with the majority of them being significantly different between the baseline (BL) and the first measurement with the damage on the cartilage of the medial compartment (MED) but not the subsequent measurements. Several parameters (duration of hits, signal strength and absolute energy) were also significantly different for the latter stage with the additional damage on the patellofemoral area (MEDPAT), and the amplitude of hits was significantly different from the BL for all measurements apart from the last one (LAT). Contrary to the LF sensor, the AE parameters registered by the HF sensor were not discernible between the different measurements. For both sensors, the b-value of the intact knee (BL) was lower than values for the following measurements with progressively damaged joint. Although this shows that the presence of cartilage damage can be detected using b-value, the extent of the damage was not. The data also contradicts previously reported notion where recovered knees exhibited lower b-values than the injured [14], and cadaveric knees with intact meniscus showed higher b-values in comparison to the knees with meniscus tears.

Overall, it can be said that for the first measurements (after 1 or 2 reopenings of the knee compartment), a notable difference in AE events number and parameters can be observed between the damaged (MED, MEDPAT) and the intact knee (BL) and it corresponds to general findings in the literature [19]. However, this trend was not maintained in subsequent measurements. It was previously reported that opening the knee compartment, as well as injecting the tissue with saline solution, does not change the recorded joint acoustic emissions significantly compared to a baseline measurement [34]. However, the knee compartment was not repeatedly reopened, and the incision used in [34] was shorter and horizontal. In this study, repetitive reopenings of the knee compartment and much larger vertical incisions were used, that potentially contributed to the overall loss of the structural integrity of the knee's soft tissues, the formation of small air pockets, while increasingly disrupting the transmission of the acoustic signal from the damaged area to the sensors' surface. Though it can be recommended for future studies on AE monitoring that are using cadaver specimens to avoid repetitive reopenings of the joint compartment, that might not be an optimal approach when considering the value of cadaveric material. Another solution to avoid the detrimental effect of reopening the incision would be to use minimally invasive approaches, e.g., the mini-medial parapatellar approach to open the knee compartment, as well as ensuring that no air pockets remained between soft-tissue layers by compressing them before the knee's closure, and adopting a suturing technique that provides an airtight seal.

B. Sensors' Frequency

The HF sensor overall showed low sensitivity, as none of the included AE hit parameters were significantly different in the conducted measurements. However, from the amplitude distribution plot (Fig. 7), it was evident that in the range between 26 to 36 dB, the baseline can be distinguished from the measurements of the damaged articulating surfaces, but those measurements cannot be distinguished from each other. It was also shown by they calculated b-values, that the same measurements carry higher values than the baseline (0.09), but close to each other (0.13, 0.13, 0.14), with the exemption of one trial (Fig. 8). Lowering the threshold for the AE hit detection further from the 24dB that was used in this study, may improve the performance of the HF sensor.

Overall, the LF sensor was able to distinguish among the initial stages across multiple parameters of AE events, such as the signal's strength and absolute energy (that were significantly different for the first two measurements), counts to peak and hit amplitude (significantly different for the first three stages of damage), despite the compromised AE transmission that potentially resulted from the repeated reopening of the joint and the damage being located further away from the sensor. With respect to the number of hits, the LF better differentiated (Fig. 5(b)) intact knee from first stage of damage (MED), which is also evident by the amplitude distribution graphs (Fig. 8) and the computed b-values (BL: 0.13, stages of progressive damage: 0.20, 0.15, 0.17 and 0.14).

C. Sensors' Position

A slightly lower number of hits per repetition and considerably fewer variable data (Fig. 9) were registered in the medial position for both sensors for measurements when the cartilage damage located in the medial compartment (MED), indicating that the lateral position is more prone to unsystematic artefacts and noise (high amplitude AE events, despite further distance from the damaged area). These findings fall in line with previous work [21] that indicates that the medial tibial condyle area of the knee shows less muscular and dynamic artefacts. A lower amplitude and average frequency was observed in the recorded signal by the LF sensor (Table II) that indicates the increased attenuation of the high frequency portion of the signal in the soft tissue, which is in line with the data previously reported in the literature on the acoustic waves transmission in soft tissues [41]. However, when registering with the HF sensor, the average frequency did not change between sensors, confirming previous findings that in the considerably higher frequency ranges (50-600kHz) the attenuation is insensitive to the frequency of the AE hit [42]. While the HF sensor is less sensitive to the AE caused by the introduced damage, this finding may be of importance in applications where a high signal attenuation is expected and it is also important to preserve the frequency distribution of the AE signal.

D. Angular Position

The recordings obtained during different stages of knee flexion (Fig. 10) can potentially be used to locate the cartilage damage, as specific articulating cartilage surfaces are in contact with each other or under load during the different phases of knee flexion. Thus, for HF sensor where in case of medial compartment damage a bigger number of hits were registered for flexion angles close to 0, whereas for next measurement with additional patellofemoral damage same movement range showed lower number of AE events. However, the same was not observed with the LF sensor, as the measurement with progressive joint damage showed generally decreasing number of AE events, potentially due to its lateral position of the sensor, which was more affected by repeated reopening of the knee compartment.

V. Limitations and Future Work

This paper presents a pilot proof-of-concept study and its limitations should be considered when interpreting the results. AE sensors with specific characteristics, specifications, and explicit detection parameters (as outlined in Section II-B) were used to record and define AE events. In this respect, such an analysis may benefit from the use of multiple and dissimilar sensors simultaneously, and the recording and analysis of the full waveform of the acoustic signals, allowing for the further frequency analyses and comparisons of the AE events associated with cartilage damage. Only one specimen was considered in this pilot study and further investigation is needed to confirm the findings, after taking into account the suggested measures to improve consistency between trials. Considering the mechanical manipulation and loading of the specimen, changing the way the knee compartment is closed between the measurements or avoiding repetitive reopenings of the knee altogether, will result in more reliable measurements.

While particular attention was paid to imitate the loss of cartilage due to osteoarthritis, the donor of the specimen was relatively young and no signs of cartilage softening or chondromalacia were observed upon the opening of the joint, which might affect the tribological properties of the cartilage and the generation and transmission of the recorded AEs as a result. Additionally, it needs to be noted that joint AEs were recorded in starved lubrication conditions (fresh frozen specimen with compromised joint synovial capsule). Thus, comparing the recorded signals with cadaveric knee specimens with natural OA damage should also be carefully considered. Specifically, in subsequent research, a larger sample size of healthy volunteers of all ages should be included, in addition to patients suffering from osteoarthritis as well as other joint conditions.

Moreover, the specimen storage and preparation techniques can also have an effect on mechanical properties of the cartilage and surrounding tissues. While it has been previously reported that storage at subzero temperatures and rapid thawing do not affect the biomechanical properties of articular cartilage [43], slow thawing under room temperature might result in decreased stiffness [44], especially when taking into account the large size of the specimen and the area of interest. While decreased stiffness of the cartilage might not have detrimental effect on the measurements in our case (young donor), as it naturally occurs with OA knees due to chondromalacia, for a specimen that is obtained from elderly or with natural OA damage, rapid thawing in warm water may be opted to avoid additional changes in tissues.

Further research in cadaveric studies might include the investigation of AEs before and after interventions aimed at improving joint lubrication (e.g., with hyaluronic acid injections) or cartilage repair (e.g., grafting). Such studies can further investigate the potential of AE monitoring to evaluate the effects of OA treatments and therapies in vivo.

VI. Conclusion

This pilot study presents an investigation on the AEs from a cadaveric knee joint during flexion/extension following progressive OA damage. During the first two reopenings of the knee compartment, a notable difference was observed between the measurements of the damaged cartilage and intact knee, and these results correspond to the general findings in the existing literature. However, with further damage inflicted on the cartilage of the articulating surfaces of the knee, this difference progressively diminished. Repeated reopening of the knee compartment in this study have potentially affected the transmission of AEs through the soft tissues of the knee. While the use of cadaveric specimens provides a unique opportunity to investigate AEs in a highly controlled setting and allows the introduction of artificial cartilage damage emulating real conditions in a specific location of the knee compartment, such tests need to be used with caution considering the state of the surrounding soft tissue, i.e., ensuring that the damage remains minimal and that the transmission of the AE is maintained. The use of the sensor with 15–200 kH range provides better differentiation between intact knee and measurements with cartilage damage, especially for the following AE event parameters: hit amplitude, signal strength, and absolute energy. Positioning the AE sensor over the medial tibial condyle is advisable, considering that this area is less prone to motion artefacts and unsystematic noise.

Appendix

Supplementary files: S1 Table.xlsx - means, standard deviations, median values and interquartile ranges for the full list of the calculated AE parameters for HF sensor. S2 Table.xlsx - means, standard deviations, median values and interquartile ranges for the full list of the calculated AE parameters for S LF sensor. S LF.pdf - statistical tests for LF sensor. HF.pdf - statistical tests for HF sensor. The dataset acquired in this study is openly available in Zenodo [45].

Acknowledgment

The authors want to thank the ASSERT Centre, College of Medicine and Health, University College Cork, for the support provided in this study.

References

[1] "Global burden of disease (GBD 2019) | institute for health metrics and evaluation," 2021. Accessed: Oct. 20, 2021. [Online]. Available: http://www.healthdata.org/gbd/2019

[2] L. Murphy et al., "Lifetime risk of symptomatic knee osteoarthritis," *Arthritis Care Res.*, vol. 59, no. 9, pp. 1207–1213, 2008, doi: 10.1002/art.24021.

[3] M. Blüher, "Obesity: Global epidemiology and pathogenesis," *Nature Rev. Endocrinol.*, vol. 15, no. 5, pp. 288–298, Feb. 2019, doi: 10.1038/s41574-019-0176-8.

[4] F. W. Roemer et al., "State of the art: Imaging of osteoarthritis—Revisited 2020," *Radiology*, vol. 296, no. 1, pp. 5–21, May 2020, doi: 10.1148/radiol.2020192498.

[5] R. Campbell et al., "Perceived benefits and limitations of using patient-reported outcome measures in clinical practice with individual patients: A systematic review of qualitative studies," *Qual. Life Res.*, vol. 1, pp. 1–24, Sep. 2021, doi: 10.1007/S11136-021-03003-Z/TABLES/2.

[6] F. Dobson et al., "OARSI recommended performance-based tests to assess physical function in people diagnosed with hip or knee osteoarthritis," *Osteoarthritis Cartilage*, vol. 21, no. 8, pp. 1042–1052, Aug. 2013, doi: 10.1016/J.JOCA.2013.05.002.

[7] E. R. Dorsey and E. J. Topol, "Telemedicine 2020 and the next decade," *Lancet*, vol. 395, no. 10227, p. 859, Mar. 2020, doi: 10.1016/S0140-6736(20)30424-4.

[8] K. M. Esquivel, "Remote rehabilitation: A solution to overloaded & scarce health care systems," *Trends Telemed. E-Health*, vol. 1, no. 1, pp. 1–19, Aug. 2018, doi: 10.31031/TTEH.2018.01.000503.

[9] R. P. Singh et al., "Internet of medical things (IoMT) for orthopaedic in COVID-19 pandemic: Roles, challenges, and applications," *J. Clin. Orthopaedics Trauma*, vol. 11, no. 4, pp. 713–717, Jul. 2020, doi: 10.1016/J.JCOT.2020.05.011.

[10] K. Guk et al., "Evolution of wearable devices with real-time disease monitoring for personalized healthcare," *Nanomaterials*, vol. 9, no. 6, May 2019, Art. no. 813, doi: 10.3390/NANO9060813.

[11] S. Tedesco et al., "A multi-sensors wearable system for remote assessment of physiotherapy exercises during ACL rehabilitation," in *Proc. IEEE 26th Int. Conf. Electron. Circuits Syst.*, 2019, pp. 237–240, doi: 10.1109/ICECS46596.2019.8965214.

[12] G. Kontadakis et al., "Gamified platform for rehabilitation after total knee replacement surgery employing low cost and portable inertial measurement sensor node," *Multimedia Tools Appl.*, vol. 79, no. 5/6, pp. 3161–3188, Feb. 2020, doi: 10.1007/S11042-018-6572-6/FIGURES/20.

[13] S. M. Bolam et al., "Remote patient monitoring with wearable sensors following knee arthroplasty," *Sensors*, vol. 21, no. 15, Jul. 2021, Art. no. 5143, doi: 10.3390/S21155143.

[14] H. K. Jeong, D. Whittingslow, and O. T. Inan, " b-Value: A potential biomarker for assessing knee-joint health using acoustical emission sensing," *IEEE Sensors Lett.*, vol. 2, no. 4, Dec. 2018, Art. no. 7001204, doi: 10.1109/LSENS.2018.2871981.

[15] S. Hersek et al., "Acoustical emission analysis by unsupervised graph mining: A novel biomarker of knee health status," *IEEE Trans. Biomed. Eng.*, vol. 65, no. 6, pp. 1291–1300, Jun. 2018, doi: 10.1109/TBME.2017.2743562.

[16] L.-K. Shark et al., "Discovering associations between acoustic emission and magnetic resonance imaging biomarkers from 10 osteoarthritic knees," *IEEE Trans. Biomed. Eng.*, vol. 69, no. 11, pp. 3494–3503, Nov. 2022, doi: 10.1109/TBME.2022.3171493.

[17] R. Unnþórsson, "Hit detection and determination in AE bursts," *Acoustic Emission - Research and Applications*, InTech, Mar. 2013, doi: 10.5772/54754.

[18] J. L. Katz, "Engineering criteria for biomaterials: Some thoughts on in situ measurements," *Biomaterials Med. Devices Artif. Organs*, vol. 7, no. 1, pp. 81–87, 1979.

[19] L. Khokhlova et al., "Assessment of hip and knee joints and implants using acoustic emission monitoring: A scoping review," *IEEE Sens. J.*, vol. 21, no. 13, pp. 14379–14388, Jul. 2021, doi: 10.1109/JSEN.2020.3045203.

[20] M. Browne et al., "Passive monitoring of knee joint condition using acoustic emission," *Orthopaedic Proc.*, vol. 98, no. 1, p. 54, 2016.

[21] T. I. Khan and H. Yoho, "Integrity analysis of knee joint by acoustic emission technique," *J. Multimodal User Interfaces*, vol. 10, no. 4, pp. 319–324, Dec. 2016, doi: 10.1007/s12193-015-0206-3.

[22] L. Spain et al., "Biomarkers for knee osteoarthritis: New technologies, new paradigms," *Int. J. Clin. Rheumatol.*, vol. 10, no. 4, pp. 287–297, 2015, doi: 10.2217/ijr.15.32.

[23] K. Wierzcholski, "Acoustic emission diagnosis for human joint cartilage diseases," *Acta Bioeng. Biomech. /Wrocław Univ. Technol.*, vol. 17, no. 4, pp. 139–148, 2015.

[24] D. K. Schluter et al., "Use of acoustic emission to identify novel candidate biomarkers for knee osteoarthritis (OA)," *PLoS One*, vol. 14, no. 10, 2019, Art. no. e0223711, doi: 10.1371/journal.pone.0223711.

[25] R. E. Andersen, L. Arendt-Nielsen, and P. Madeleine, "A review of engineering aspects of vibroarthography of the knee joint," *Crit. Rev. Phys. Rehabil. Med.*, vol. 28, no. 1/2, pp. 13–32, 2016, doi: 10.1615/CritRevPhysRehabilMed.2016017185.

[26] S. C. Abbott and M. D. Cole, "Vibration arthrometry: A critical review," *Crit. Rev. Biomed. Eng.*, vol. 41, no. 3, pp. 223–242, 2013, doi: 10.1615/CRITREVBIOMEDENG.2014010061.

[27] L.-K. Shark, H. Chen, and J. Goodacre, "Knee acoustic emission: A potential biomarker for quantitative assessment of joint ageing and degeneration," *Med. Eng. Phys.*, vol. 33, no. 5, pp. 534–545, Jun. 2011, doi: 10.1016/j.medengphy.2010.12.009.

[28] H. J. Schwalbe et al., "Non-destructive and non-invasive observation of friction and wear of human joints and of fracture initiation by acoustic emission," *Proc. Inst. Mech. Eng. H: J. Eng. Med.*, vol. 213, no. 1, pp. 41–48, 1999, doi: 10.1243/0954411991534799.

[29] K. Kalo et al., "Reliability of vibroarthrography to assess knee joint sounds in motion," *Sensors*, vol. 20, no. 7, Apr. 2020, Art. no. 1998, doi: 10.3390/s20071998.

[30] D. M. Hochman et al., "A pilot study to assess the reliability of sensing joint acoustic emissions of the wrist," *Sensors*, vol. 20, no. 15, Jul. 2020, Art. no. 4240, doi: 10.3390/S20154240.

[31] G. C. Ozmen et al., "A novel accelerometer mounting method for sensing performance improvement in acoustic measurements from the knee," *J. Vib. Acoust.*, vol. 143, no. 3, Jun. 2021, doi: 10.1115/1.4048554.

[32] L. Khokhlova et al., "Motion artifact resistant mounting of acoustic emission sensors for knee joint monitoring," in *Proc. IEEE 43rd Annu. Int. Conf. Eng. Med. Biol. Soc.*, 2021, pp. 7300–7303, doi: 10.1109/EMBC46164.2021.9629954.

[33] L. Khokhlova et al., "Test-retest reliability of acoustic emission sensing of the knee during physical tasks," *Sensors*, vol. 22, no. 23, Nov. 2022, Art. no. 9027, doi: 10.3390/S22239027.

[34] D. C. Whittingslow et al., "Acoustic emissions as a non-invasive biomarker of the structural health of the knee," *Ann. Biomed. Eng.*, vol. 48, no. 1, pp. 225–235, Jan. 2020, doi: 10.1007/s10439-019-02333-x.

[35] L.-K. Shark, H. Chen, and J. Goodacre, "Discovering differences in acoustic emission between healthy and osteoarthritic knees using a four-phase model of sit-stand-sit movements," *Open Med. Inform. J.*, vol. 4, pp. 116–125, Jul. 2010, doi: 10.2174/1874431101004010116.

[36] S. Cristea et al., "Surgical approaches for total knee arthroplasty," in *Arthroplasty - A Comprehensive Review*. London, U.K.: IntechOpen, 2016.

[37] T. F. Besier et al., "Patellofemoral joint contact area increases with knee flexion and weight-bearing," *J. Orthopaedic Res.*, vol. 23, no. 2, pp. 345–350, Mar. 2005, doi: 10.1016/J.ORTHRES.2004.08.003.

[38] PAC, "PCI-2 based AE system user's manual rev 3 associated with: AEWIN for PCI-2 software part #: 6301-7001 version 1.30 or higher," Physical Acoustics Corporation, Princeton Junction, NJ, USA, 2007.

[39] M. Båth, B. Gutenberg, and C. F. Richter, "Seismicity of the earth and associated phenomena," *Tellus*, vol. 2, no. 1, pp. 68–68, Feb. 1950, doi: 10.1111/J.2153-3490.1950.TB00313.X.

[40] A. Ghasemi and S. Zahediasl, "Normality tests for statistical analysis: A guide for non-statisticians," *Int. J. Endocrinol. Metab.*, vol. 10, no. 2, pp. 486–489, Apr. 2012, doi: 10.5812/IJEM.3505.

[41] T. D. Mast, "Empirical relationships between acoustic parameters in human soft tissues," *Acoust. Res. Lett. Online*, vol. 1, no. 2, pp. 37–42, Nov. 2000, doi: 10.1121/1.1336896.

[42] T. M. Wright and J. M. Can, "Soft tissue attenuation of acoustic emission pulses," *J. Biomech. Eng.*, vol. 105, no. 1, pp. 20–23, Feb. 1983, doi: 10.1115/1.3138379.

[43] A. Changoor et al., "Effects of refrigeration and freezing on the electromechanical and biomechanical properties of articular cartilage," *J. Biomech. Eng.*, vol. 132, no. 6, Jun. 2010, Art. no. 064502, doi: 10.1115/1.4000991.

[44] E. A. Kennedy, D. S. Tordonado, and S. M. Duma, "Effects of freezing on the mechanical properties of articular cartilage," *Biomed. Sci. Instrum.*, vol. 43, pp. 342–347, 2007.

[45] L. Khokhlova et al., "Data from: Non-invasive assessment of cartilage damage of the human knee using acoustic emission monitoring: A pilot cadaver study," *IEEE Trans. Biomed. Eng.*, 2023, doi: 10.5281/zenodo.7731693.